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Development of a Hemodynamics Computer Model of Human Tolerance to +Gz Accelerations

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THIS TECHNICAL REPORT HAS BEEN REVIEWED AND IS APPROVED FOR PUBLICATION.

FOR THE DIRECTOR

//signed//

MARK HOFFMAN, Acting Chief
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CONTENTS

LIST OF GENERAL SYMBOLS AND ABBREVIATIONS

Symbols used for model of heart pump function

$i=1$ low index that defined property to the right heart

$i=2$ low index that defined property to the *left* heart

H_a^V distance between atrical and venticular centers

N^G coefficient of gravitational overload

φ gravitational vector deflexion angle

P_i^{FD} end - diastolic pressure

ρ, A, B approximation constants

Q^i, O^o input and output flows respectively

r_1, r_2 resistances of opened and closed valve respectively

ΔP_k pressure difference on both valve sides

P_{KP} pressure necessary for valve opening

P_{KO} pressure necessary for valve closing

T_L duration of diastole phase

C diastolic elasticity of ventriculae

V strained volume

F heart rate

k inotropic coefficient of ventriculae.

Symbols used for model of hemodynamics in multicompartmental ramified vascular net.

P_j^I, P_j^E intra - and extra - vascular pressures respectively

P_j^T transmural pressure $P_j^T = P_j - P_j^E$

U_j non - strained blood volume

V_j sectional blood volume

D_j volumal rigidity

R_{jl} hydraulic pressure between j - th and l - th compartments

q_{jl} their blood flow

δ_M sensitiveness of brain flow selfcontroling mechanism

Symbols used for models of control mechanisms of hemodynamics

P^C mean pressure in carotid arteries

K_j^O, K_j^I output and input coefficients of compartment orientation

φ_j angle between j compartment and horizontal

h_j length of compartment

ρ_j density of blood

P_j^G gravitational component of pressure

K^E coefficient of transmission P^E into vessel

F_j^R level of mechanoreceptor activity

P_j^R transmural pressure

P_j^t thresholds of receptors activity

X_j control parametres of CVS

E_v, E_s activity of N.vagus and N.sympathicus

F_A, K_A levels of F and K under automatic regime of heart

a_i, b_i, p_j constants

d_1, d_1^u, β constants

P^I innere (tissue) extravascular pressures in each section of anti - G suit

P^E air pressure in sections of anti - G suit

P_m mean muscle pressure

g_m^{th} threshold of acceleration for muscle stress start

g_T^{th} threshold of acceleration for anti - G suit initiaton

g_{BP}^{th} threshold of acceleration for start of breathing pressure increase

P_m mean muscle pressure

P_{p_0} intrapleural pressure under human supine position

P_p intrapleural pressure under other human positions

α_{BP} gradient of breathing pressure increase

List of abbreviations:

MMC mathematical model complex

CBSC computer based software complex

CVS cardiovascular system

HPF heart pump function

CVN cardiovascular net

ABR arterial baroreceptor reflex

SCIT special computer informational technology

INTRODUCTION

The current general knowledge concerning human hemodynamics under piloting accelerations mainly based on direct observations during different profile and duration accelerations in real flight or their simulation using centrifuges. However, because of lack all of necessary information about important characteristics of human cardiovascular system such kind of observations are essentially limited to imagine of full physiological mechanismes, that usually are involving in general natural protection and are providing the human tolerance to extremaly high level of piloting accelerations. Therefore, to make wide and deeper our notions about these mechanisms also were used physiologic researches on a different animals with directly measures additional physiologic parameters [9,10]. The third direction of investigators' efforts was a creation of adequate mathematical models, which allow estimate the relative role of different control subsystems of organisms [11,13,15] in general acceleration stress. In this direction were created as relatively simple models for analytic way analyze [7,8], and much more complicate models that might include hundreds of differential equations and thus are requiring for theirs analyze computer [12,13,15].

The work presented in this report related to the last class of models. The exclusive side of this development is that in fact it is the first attempt in World practice when was aimed to create both a special problem-oriented mathematical model complex and a special software oriented to physiologist-researcher to solve by theoretically way practically all of problems which might be arise during development and test of new methods of human protection under rapid and gradual accelerations.

This report describes developed original mathematical models and also gives the information necessary for user (physiologist) to let him to be able to do computer simulation experiments. As to inside structure of computational algorithms and software, they are not described here because of such an information was not required in contract by customers of development.

According to the general conception of development and use of special computer based technology for simulation research of human hemodynamics under dynamic piloting accelerations +Gz, the mathematical model complex (MMC), that able to describe practically all aspects of investigating processes, has been created. It includes the following models:

- two models of heart pump function: one for presenting all of well known physiological regularities of heart function (heterometric and homeometric mechanisms of their self-control) for its quasi-stationary and stationary regimes, and another — exclusively for representing of dynamic processes within every cardiocycle;
- model of human systemic and lung hemodynamics under several situations characteristic for study in aviation physiology and medicine: postural changes, +Gz accelerations without or with different protections, applications of local positive or negative pressures;
- model of regulators of cardiovascular system (CVS), that contains autonomous blocks of central nervous and some of humoral control mechanisms of heart function and systemic hemodynamics.

The basic algorithms for computer realization of MMC for several defined situations as well as the main instructions for users of our software also are presented in report.

We would like to note that some useful information about potentials of our developments was included into the final variant software to show the possible ways for future developments. We hope our collaboration will be continued and this basic development will become to its final variant as one useful additional research tool for investigators.

Chapter 1. GENERAL CONCEPTION OF DEVELOPMENT AND USE OF *MATHEMATICAL MODELS*

According to general aims of this development we had to create mathematical models that able to be realized as an autonomous software for IBM compatible computers and have to simulate the physiology of human hemodynamics under gradual and rapid accelerations Gz for their wide range of changes.

To reach this general goal we had to perform the following:

- to create one basic mathematical model complex that includes both cardiovascular system's (CVS) physiology and main hemodynamic effects of all natural and artificial tools and methods used to provide of human tolerance to the high level Gz accelerations;
- to choose the appropriate algorithms for computing of created systems of quantitative equations;
- to create an optimal software for these computations;
- to find out all necessary values for constants of models and to reach acceptable simulation results for all known tests situations;
- to create one special user interface oriented to the physiologist-researcher and an ability to provide him computer simulation experiments and their results' analyze in easy way.

The first item of the list mentioned above is subdivided into two branches: first- what kind of mathematical models could be able to adequately describe the human hemodynamics in analyzed conditions, and second- what type of relations must be estimated as sufficient when we want to describe the transformation process of physical efforts to hemodynamic effect. This problem relates to modeling of protective tools (anti-G suits chambers inflantation) or human conscious acts (for examples, muscle stressing, special breathing regimes). The practice of hemodynamics modeling [11,12,15] gives us some well based approaches about the first problem. But we could not find any example of successfully solution for the second problem. Taking into account these aspects [11,12], we had chosen one of the most known approaches of modeling, namely the approach of multicompartmental models, for simulation of central hemodynamic effects during accelerations.

As to models of exsogenic (relatively to the CVS) factors, we have used relatively simple mathematics (differential or algebraic equations with constant coefficients). The chosen approach has allowed us to describe conceptually the main biophysical processes. After a lot of computer experiments we became sure, that we have got quite right values of models' constants for analyzed class of situations.

Taking into account the reasons mentioned above we have proposed the general concept of development and use of computer based software complex (CBSC). This concept is shown on fig.1.

According to this conception, it is enough to have one basic mathematical model complex (MMC), that could be actualized by user and prepared for every simulation experiment with a special collection of experimental parameters. He/she could also be able to choose the scenario of actual simulation experiment, to start it and to interrupt or break it (when it is necessary). At the end of simple computer experiment the user will be able to analyze its results. We proposed two form for this analysis: graphic and table. We also propose to save the result of current experiment and to compare graphic results of two different experiments when it is interesting for researcher.

The MMC consists of two different parts, that we call endogenic and exogenic parts (see fig.2). The endogenic part includes a model of heart pump function, a model of hemodynamics in ramified vascular net (these two models are able to describe hemodynamics under constant level of central nervous activity), and a model of nervous control mechanisms of CVS. These three models together describe the human hemodynamics for his/her horizontal or seated positions. Every transitory process in hemodynamics under human postural changes will finally bring the hemodynamics to a new stable regime. Time intervals necessary for transitory processes between two different stable states are usually less than one minute.

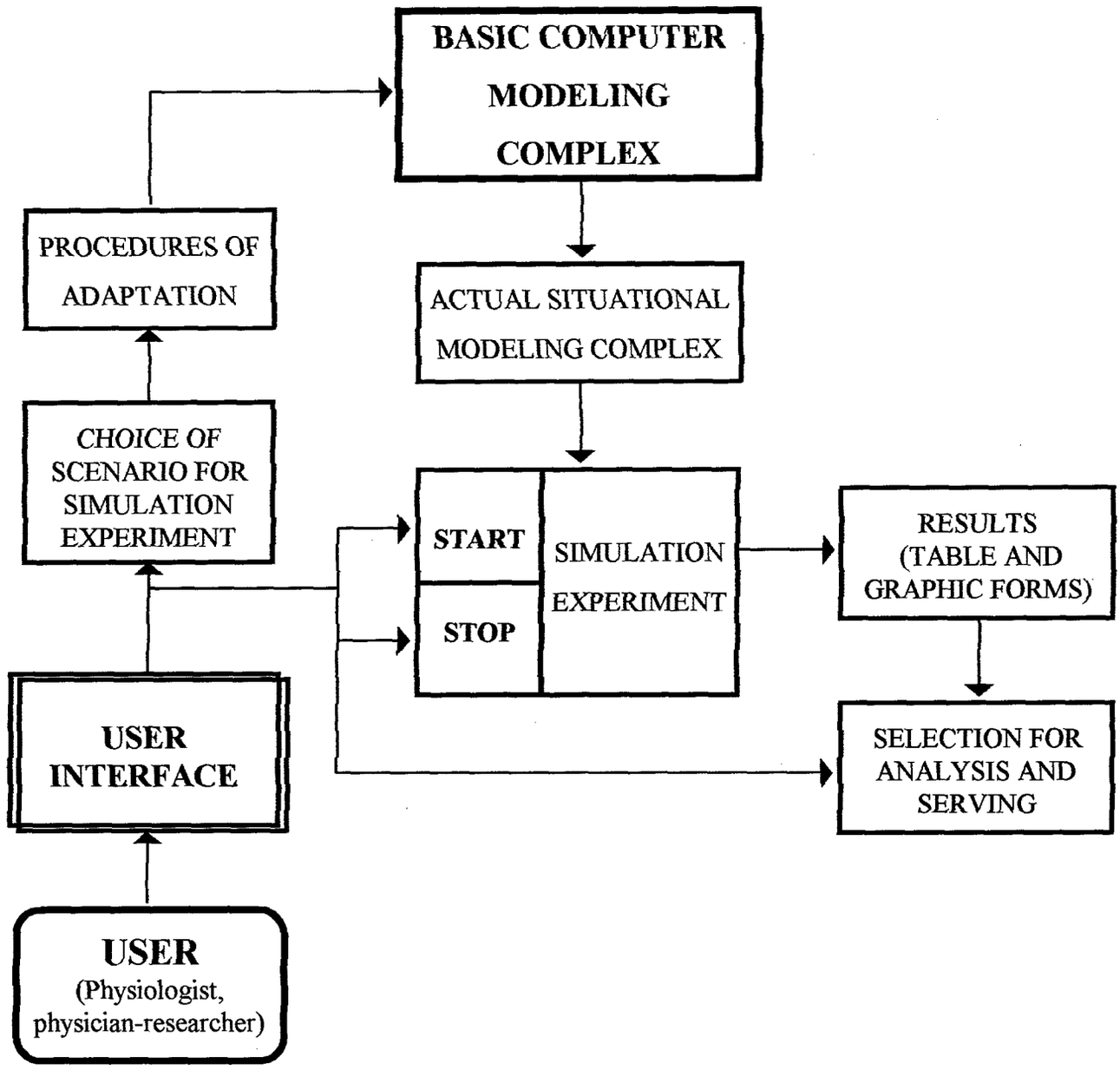


Fig.1. General conception of use of computer based mathematical models.

The endogenic parameters of models are as follows:

- the summary blood volume in CVS;
- the background level of heart rate (HR) for human clinostatic, upright seated or erect positions;
- the level of cathecholamines' concentrations in blood. This level defines the general stress. It might be indicated by HR value just before onset of accelerations;
- the ratio between power of baroreflexes from aortic arch and carotid sinus area;
- the ratio between baroreceptor reflex control of heart pump function parameters (HR and inotropism) and parameters of vessels' tonus, including regional differences of arterial and venous compartments;
- the value for every time constant for different baroreceptor and mechanoreceptor reflexes;
- the sensibility of heart and vascular effectors to the level of efferent nervous activity;
- the reserve for all CVS's control parameters;

The user could directly set some of these parameters using software interface. For more detailed information about such possibility see chapter 4.

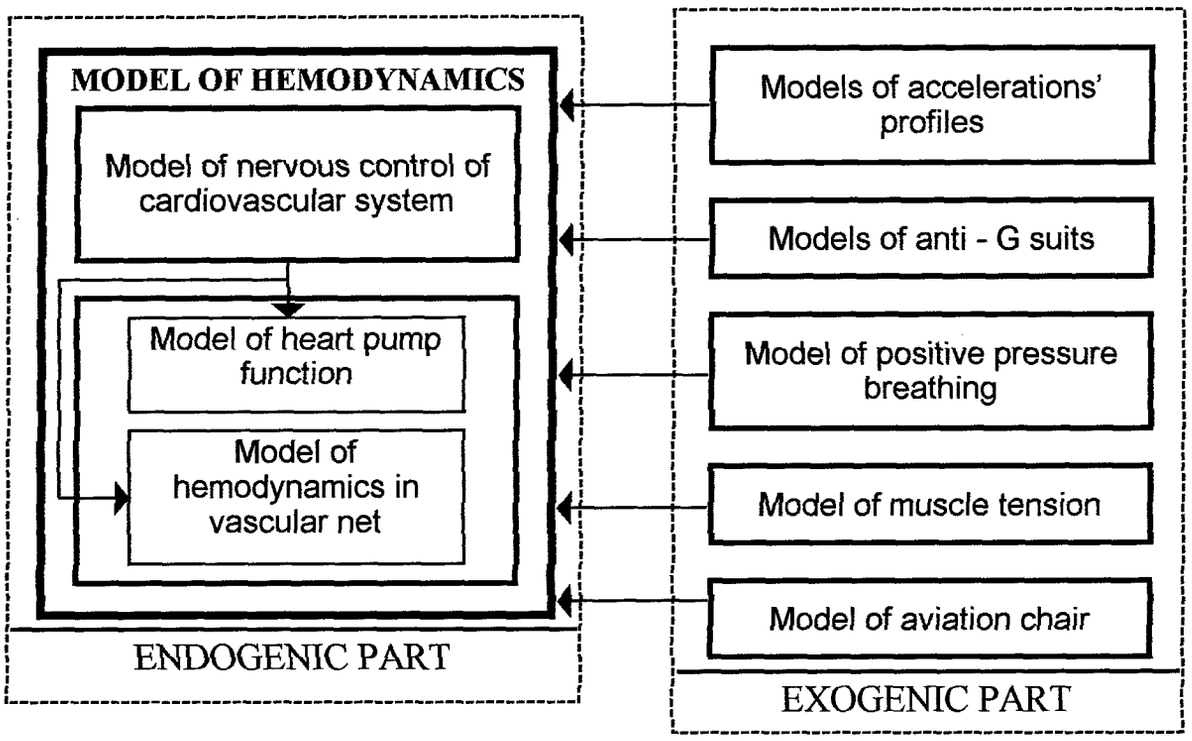


Fig.2. The inside structure of basic model complex.

The exogenic part of MMC includes five different models. They all are related to the mechanisms relatively external to the CVS. These models are necessary to describe hemodynamic reactions under external loading (by different profile and duration accelerations) both for human fixed position on an armchair and during changes of several angles of body parts relatively to the acceleration vector's direction.

Models of accelerations describe several fixed acceleration profiles (in this variant we present 4 fixed profiles. Every of these profiles have its several parameters which might be set by user) and an arbitrary profile, which might be special constructed by user.

The model of armchair allows to change three angles of autonomy body parts: calf (shank), thigh and remained upper body part (head, thorax and abdomen). These angles must be set before to start the simple simulation experiment.

The model of muscle tension describes the dynamics of extravascular pressures for every vascular compartment. Such changes of muscle tensions might be initiated both consciously and unconsciously by simple reflector way's activation during postural changes .The software proposes users three different levels of muscle stress. It is important to note here that this model is dynamic. Therefore the user will be able to set both the threshold of acceleration for start increasing of muscle tension and the maximal level of this tension as a extravascular pressure.

The models of anti-G suits are also dynamic and describe the value of extravascular pressure for all vascular compartments located in areas of abdomen, thigh and shank parts of legs. There are two setting parameters for each of these parts: the level of acceleration (threshold) to start of appropriate chambers' inflantation and the gradient of pressure's increasing.

The model of positive pressure breathing describes the dynamics of pleural and abdominal pressures during different breathing regimes. Here also we propose users to set both the G-level thresholds to start this protective procedure and the gradient of breathing pressure's linear increasing for every G. Besides, the model proposes to set the top-limit for positive pressure value.

All above-mentioned sentences are concerning to the healthy person. It should be noted that the customer also asked us to analyze the problem of creation some of special models for several diseases. Customers proposed the list of these diseases and we have worked on this problem. But some later it became clear that to create adequate models for these additional situations we need

much more special information about dynamics of transformation the healthy man to the weak one. Such kind of information was absent. So, we have done only some first attempts in this direction and have developed relatively simple models. These models are included in general CBSC and are presented in this report. But we cannot guarantee their adequacy. We are sure that this aspect is very interesting, but at the same time it should be considered as a special problem and might be specially solved in another development.

Chapter 2. GENERAL INFORMATION ABOUT BASIC MATHEMATICAL MODELS

2.1. Models of Heart Pump Function

We had two starting conditions for creating the basic MMC and for its further transformation into an appropriate research technology oriented towards the area of human acceleration physiology. According to the first condition, our models must be able to describe almost all known physiologic mechanisms that may be involved by human healthy organism in general activation process to reach some level of human tolerance to dynamic high sustained +Gz accelerations. According to the second condition, the required models must also be able to reflect the human hemodynamics under the same situations but including several additional conditions that have been presented later by customers as a list of seven special situations:

1. Mitral valve prolapse;
2. Aortic insufficiency;
3. Mitral valve regurgitation;
4. Ventricular tachycardia;
5. Supraventricular tachycardia;
6. Bradicardic arrhythmia;
7. Push-pull effect during aerial combat maneuver.

Six situations of this list are related to the different cardiac pathology. One group of them (four situations) may appear by some dissotiations between contraction-relaxation rhythms of different heart chambers. Besides, there are also two situations with defects of heart valves.

These six situations are rather hard to realize in frame of one simple quantitative mathematical model. Therefore we must create an adequate special model complex and, at the same time, we should include it in a frame of one computer technology, convenient for physiologist-researcher and relatively simple in use.

In our opinion, the seventh situation may be considerate as a peculiarity of behavior of hemodynamics rather than a special model category. In fact, all these peculiarities will be caused by dynamics of postural changes of pressures, flows and blood volumes in body different parts. Therefore, we have reasons to expect that our models will be able to handle this specific effect automatically.

So, it was clear that a special problem-oriented model complex should be developed. It will include several necessary autonomous and relatively simple models. To meet these requirements we intend to use two models of heart pump function (HPF). The first model of HPF is considered as the main one for all normal physiological situations, when the second model of HPF should be used only during simulation of cardiac diseases. Such combination of HPF-models gives us an opportunity to simulate human hemodynamics without reflection of details for dynamics of pressures, flows and volumes within each cardiocycle, if we do not need it. Using this form of simulation we will be able to operate only with the mean values of control parameters for each cardiocycle..

There have been several reasons to consider the second model of HPF. First of all it does not require too small discretization of step for approximate solution of general system of integro-differential equations of model complex for the whole CVS. Thus we can essentially economize on the calculation time. Therefore we can be sure that our computer simulator will be able to provide users with simulation regimes similar both for a real piloting maneuvers and for regimes that take place during real time simulation of accelerations by centrifuge.

But at the same time, if we want to simulate all listed regimes of cardiac pathologies mentioned by customers, we will be compelled to describe the dynamics of intracardial hemodynamics for each of heart chambers. As our preliminary investigations in this aspect of simulation have shown, the biophysical appropriate and mathematically adequate computer model of

HPF demands much more detailed computing (calculations), than the first model of HPF mentioned above.

On the one hand it might seem that we are speaking about some inside details of modeling technology. But they are really important to discuss them specially. To make clear how important these inside problems of computer modeling may be, it is reasonable to turn reader's attention to several aspects of computing technology. The whole model complex will be presented as a rather large system of algebraic-differential equations (This system includes more than 1000 equations). Such system of equations may by solved only using approximate methods. The exactness for equation solution in the framework of such mathematical technology depends on the value of discretization step (H) of this process. But on the other hand it is well known, that the decreasing of H evokes the lengthening of calculation process. We know that the calculations for the first model are acceptable when H=0.01s. In the frame of second model H will be <= 0.001s to provide the fluent and stable calculations.

To provide the function of integral model complex, including the model of CVS based on two different HPF-models and models of special environment for fighter pilot, an appropriate complex computing algorithm has been developed. It is considering two ways of simulation of HPF. Besides, our algorithm automatically chooses the appropriate HPF- model and exclusive value of H for calculations into HPF-block whereas the value of H for the whole model of CVS can be different. An additional information about the proposed computing algorithm is presented in the corresponding part of this report (Chapter 3).

2.1.1. Basic model of heart pump function.

Taking into account conditions and limits described above, the main model of HPF discloses main relations between the mean values of cardiac output (Q) and central venous (Pv - for right heart) or lung venous (Plv - for left heart) pressures. Additional factors, that have been taken into consideration, are the heart rate (F) and the inotropic coefficient (k) of ventriculae (k reflects almost the linear relation between ventriculae's end-diastolic volume and stroke volume for the wide changing range of input venous pressure), the resistance (Rav) of atria-ventricular valves, the diastolic elasticity (C) and the unstressed volume (U) of ventriculae, the duration of diastole (T):

$$
\begin{aligned}
Q &= \frac{F \cdot k_i \cdot \left[\left(\Delta P_{ai}^{V} \cdot C_i + U_i\right) - U_o\right] \cdot \left[1 - exp\left(-\frac{T_L}{R_{aVi}^{K} \cdot C_i}\right)\right]}{1 - \left(1 - k_i\right) \cdot \left[1 - exp\left(-\frac{T_L}{R_{aVi}^{K} \cdot C_i}\right)\right]} \\
P_{ai}^{V} &= P_i^{INP} + 0.73 \cdot \rho \cdot H_a^{V} \cdot N^{G} \cdot \sin\varphi - P_i^{FD} \\
T_L &= \frac{1}{F} \cdot A + B \cdot \left(1 - k\right) \\
P_i &= \begin{cases} 0, & V_i \le U_i \\ \dfrac{\left(V_i - U_i\right)}{C_i}, & V_i > U_i \end{cases} \\
V_i(t) &= V_i(0) + \int_0^t \left(Q^{I} - Q^{O}\right) dt \\
R_{aV}^{K} &= \begin{cases} r_1, & \Delta P_K > P_{KP} \\ r_2, & \Delta P_K \le P_{KO} \end{cases} \\
V_i^{s} &= k \cdot V_i^{ed} - U_0
\end{aligned}
\qquad (2.1)
$$

The last formula in (2.1) is another reflection level of well-known regularity of HPF. This regularity is also known as the heterometric or Frank-Starling's mechanism of self-control of heart function. It shows that relations between stroke volume (V_i^s) of ventriculae and their end-diastolic volume (V_i^{ed}) may be presented as a linear approximation. The coefficient *k* here characterizes the level of inotropism of heart. Therefore, using *k* we can establish relations between influence of concentration of cardioactive humoral agents or rate of efferent nervous activity and inotropic state on the right or left ventriculae of heart .

2.1.2. Model of heart pump function for simulation of several cardiogenic diseases.

The second model of HPF included in the model complex describes main relations between pressures, volumes and flows (as output characteristics of heart) and dynamics of some biophysical parameters for each of heart chambers within a simple cardiocycle depending on levels of input (venous) and output (aortal) pressures. The list of parameters includes end-diastolic and end-systolic levels of capacities, considering forms of their transitions during changes of cardiac phases from diastole to systole and in the opposite direction, also hydraulic resistance of valves. Using this model we can set different values for each of heart valves, the different frequency of relaxation-contraction cycles for each atria or ventricula and thereby to simulate the human hemodynamics under +Gz accelerations for all regimes, mentioned in the customer list of diseases.

To simulate characteristic forms of pulsatile behavior of hemodynamic parameters within simple cardiocycle, it is necessary at least to find the appropriate and convenient mathematical descriptions for intracardial dynamics of myocardial rigidity (D) both for diastolic relaxation and for systolic contraction phases. Our previous investigations in this area had shown that dynamics of these rigidities for each of heart chamber are similar to the direct (into the phase of systolic contraction) or to the overturned (into the phase of diastolic relaxation) *S*-form curves. If we know the end-diastolic (D^{ed}) and the end-systolic (D^{es}) values for each of heart chamber, then the formula (2.2) lets us describe cardiocyclic dynamics for D (t) in the following compact form:

$$D_i(t) = \begin{cases} D_i^{ed} + \left(D_i^{es} - D_i^{ed}\right) \times \left[\left(\exp\left(-\alpha_i \times T\right) - 1\right)\left(\beta_i + \exp\left(-\alpha_i \times T\right)\right)\right]^2, \; i = \overline{1,4} \text{ and } \left(0 < T < t1\right) \\ D_i^{ed} + \left(D_i^{es} - D_i^{ed}\right) \times \exp\left(\gamma_i \times T\right), \quad \begin{array}{c} i = 1,3 \text{ and } t_2 \le T \le T_c \\ i = 2,4 \text{ and } \left(0 \le T < t_2 \text{ or } t_3 \le T < T_c\right); \end{array} \end{cases} \tag{2.2}$$

where i = 1 - right atria, i = 2 - right ventriculae, i = 3 - left atria, i = 4 - left ventriculae, T - time within every cardiocycle, $\alpha i, \beta i, \gamma i$– are constants that characterize the functional state for each of heart chamber. These constants may have different values depending on organism's general neuro-humoral activity.

The formula for pressure calculation is analogous to the formula described above in (2.1) and looks like:

$$P_i(t) = D_i(t) \left[V_i(t) - U_i(t) \right], \text{ for } i = 1,2,3,4 \tag{2.3}$$

Only after describing the heart valves' function it is possible to simulate the dynamics of directed flow and outflow for each of heart chamber as a sequence of periodic changes in rigidity of myocardium.

Function of atriaventricular valves for right and left hearts are described in the model by analogous mathematical expressions: as two possible states of their hydraulic resistance (R_v), depending on the pressure gradient (G_r) between atrial (P_1) and ventricular (P_2) sides of valve ($G_r = P_1 - P_2$). The real dynamics of valve transitions in direct or in opposite directions were omitted. The function $R_v(t)$ is described only as a discrete transition from valves' full-closed state to full-opened state and vice versa. The relatively simple formula looks like:

$$R_v = \begin{cases} R_0, & G_r > P_{k0} \\ R_c, & G_r \le P_{kc} \end{cases}, \qquad (2.4)$$

where P_{ko} and P_{kc} characterize critical pressure gradients necessary to open or to close the valve.

Our approach to simulate the aortal valve function differs from the approach described above. It reflects also some peculiarities of valve dynamics, including oscillatory process of valves in the final phase of cardiac systolic phase. We hope that such model may be useful for analyzing the role of violation of normal function of aortic valves and for simulating appearances of some regurgitation under high-sustained +Gz accelerations.

$$R_{vai} = \begin{cases} R_{vai}^{max} - \xi_i \times \exp(\psi_i \times \Delta P_{iv}) \times \sin(\omega_i \times \Delta P_{iv}), & \Delta P_{iv} \le 0, \\ R_i^0 + (R_{vai}^{max} - R_{vai}^{min}) \times \exp(-\eta_i \times \Delta P_{iv}), & \Delta P_{iv} > 0, \end{cases} \qquad (2.5)$$

Dynamics of blood volumes in each of heart chamber is calculated by analogy with the corresponding formula in (2.1).

2.2. Problem-oriented model of human hemodynamics in multicompartmental ramified cardiovascular net.

The systemic arterial pressure (Pa) increases essentially under piloting acceleration +Gz. This increase is almost proportional to the acceleration level. Pa may be reached about 300 mm Hg and more under high-sustained accelerations. The additional (hydrostatic) increase (or decrease, when the direction of acceleration vector is opposite to the flows direction) of peripheral arterial pressure is proportional to the distance between aortic arch and place of localization of this peripheral zone. This rule acts also for peripheral venous pressures. The only difference in the last case is that the absolute level of central venous pressure does not be essentially changed from its level for the human sitting position. So, the arterio-venous difference remains approximately invariable on the same levels of vessels' localization.

We know that the biophysical pressure-volume (P-V) static characteristics of different arterial or venous vessels are essentially nonlinear. Therefore, only taking into account this fact while we are modeling of hemodynamics in CVN we can hope that during the simulation an observed shifting of blood volumes towards the direction of accelerations will be similar to the real one.

Basic characteristics for modeling of hemodynamics in ramified CVN are the static nonlinear dependencies between transmural pressures and blood volumes in each i-th compartment of vessels. These dependencies are different for different arterial or venous vessels. These nonlinear curves are approximated in the model by means of piecewise-linear characteristics, consist of three parts. According to this approximation, typical description of P-V- dependency looks like:

$$P_i^T = \begin{cases} (V_i - U_i) \cdot D_{0i}, & V_i < U_i \\ (V_i - U_i) \cdot D_{1i}, & U_i \le V_i \le U_{1i} \\ (U_{1i} - U_i) \cdot D_{2i} + (V_i - U_i) \cdot D_{1i}, & V_i > U_{1i} \end{cases} \qquad (2.6)$$

Blood flows between j-th and l-th vessel compartments, which are connected by means of hydraulic resistance R_{jl}, are defined as a result of division of pressure gradients (G_{jl}^P) by R_{jl}. *Transmural pressures, external pressures* P^E *and hydrostatic pressures* P^G *are considered as factors,* determining G_{jl}^P. Coefficients for P^E reflect differences for levels of vessels' location and transmission characteristics of different vessels environment (muscles, cavities, skin):

$$
\begin{aligned}
q_{jl} &= \frac{G^{P}_{jl}}{R_{jl}} \\
G^{P}_{jl} &= \left(P^{T}_{j} + K^{e}_{j}\cdot P^{E}_{j} + K^{O}_{j}\cdot P^{G}_{j}\right) - \left(P^{T}_{l} + K^{e}_{l}\cdot P^{E}_{l} + K^{I}_{l}\cdot P^{G}_{l}\right) \\
K^{O}_{j} &= \begin{cases} 1, & sin\varphi_j > 0 \\ 1, & sin\varphi_j > 0 \end{cases} ;K^{I}_{j} = 1 - K^{O}_{j} \\
P^{G}_{j} &= N^{G}\cdot\rho\cdot h_j\cdot sin\varphi_j
\end{aligned}
\qquad (2.7)
$$

For collapsible vessels we use special formulae to define values of their resistances:

$$
\begin{aligned}
R_1 &= \begin{cases} R_0\cdot\left(\frac{V_0}{V}\right)^2 & ,P^T > P_0 \\ R_0\cdot r_0^4\cdot\frac{a^2+b^2}{2a^3\cdot b^3} & ,P_1 \le P^T \le P_0 \\ R_1 & ,R_1 >> R_0,\ P_1 < P_0 \end{cases} \\
a &= \frac{V\cdot r_0^2}{V_0\cdot b} \\
b &= \frac{1}{3}r_0\cdot\left[d + 2\cdot\left(1+\sqrt{1-2d^2+d}\right)\right] \\
d &= \frac{V}{V_0} \\
R &= R_u\cdot\left(\frac{U}{V}\right)^2 \\
V_0 &= V\big|_{p=0}
\end{aligned}
\qquad (2.8)
$$

It is well known that the summary brain flow remains almost stable or even it may get some increase under moderate +Gz accelerations. There are two main causes of such its characteristic behavior. The first one is the self-controlling mechanism which practically provides independence of summary brain flow during essential changes of arterial (input) pressure. This alteration interval equals approximately 100 mm Hg. Outside of this interval the dynamics of resistance of brain arteiolae has another character. In fact, this mechanism has nervous origin with the time constant δ_M about 5 s. The second mechanism of the brain flow stability caused both by simultaneous changes of input-output pressures and pressure changes in extravascular (cerebral) liquid. Our model includes description of both these mechanisms as:

$$
\begin{aligned}
R^{AM} &= \begin{cases} R^{AM}_{min} & ,P^{AM} \ge P^{AM}_{max};\ P^{AM}_{min} < P^{AM} < P^{C}_{min} \\ R^{AM}_{max}\cdot C & ,P^{C}_{max} < P^{AM} < P^{AM}_{max} \\ \frac{E_1}{P^{AM}} & ,0 \le P^{AM} \le P^{AM}_{min} \end{cases} \\
&\text{where} \\
C &= \left[1 - exp\left(X_i\cdot\left(P^{AM} - P^{AM}_{max}\right)\right)\right]
\end{aligned}
\qquad (2.9)
$$

$$
\frac{dR^{AM}(t)}{dt} = \frac{\delta_M\cdot P^{AM}(t) - R^{AM}(t)}{T_m}\ ,\ P^{C}_{min} < P^{AM} < P^{C}_{max}
$$

Function of systemic venous valves is described in models as:

$$R_{jv} = \begin{cases} R_{1j} & ,q_j > 0 \\ R_{2j} & ,qj \le 0 \end{cases} \quad ,R_{1j} >> R_{2j} \tag{2.10}$$

The second variant of modeling of hemodynamic effects of acceleration is based on the calculation of every hydrostatic pressure as a function of human posture and value of acceleration. We have two classes of compartment level. The first one characterizes the value of distance between human foots and place of localization of vessel compartment for human clinostatic or erect positions. The second class (we call it real levels) reflects the value of hydrostatic pressures of human vessel compartments for all other his positions. Using values of angles between horizontal and directions of different body parts (α – for calf, β – for tigh, and γ – for all other compartments of body and head vessels), these parameters can be calculated in the model according to the following formulae:

$$\begin{aligned} L^s &= 0.5 \times A \times l^s \times sin\alpha \\ L_1^t &= \left(L_c - 0.5 \times l_1^t \times sin\beta\right) \times A \\ L_2^t &= L_1^t - 0.5 \times A \times l_2^t \times sin\beta \\ L_p &= L_c \times A \\ L_{li}^b &= \left(l_{li}^b - L_0\right) \times A \times sin\gamma + L_0 \end{aligned} \tag{2.11}$$

where l^s – length of calf, l_1^t - l_2^t – lengths of two parts of tigh, L_c – summary length of legs, L^s – level of shank vessel compartment, L_j^t and L_2^t – real levels of tigh vessels compartments, L_p – level of aviation armchair seat place, L_0, L_{ji}^b, l_{ji}^b – real level and initial length of localization for each i-th body or head vessel compartment.

Dynamics of blood volumes in compartments are described by the following equations:

$$\begin{aligned} V_j(t) &= V_j(0) + \int_0^t \left(q_j(t) - q_1(t)\right) dt \\ \sum_i V_i(t) &= \text{const} \end{aligned} \tag{2.12}$$

According to the last equation the summary blood volume in CVS is stable during the relatively short time interval of simulation experiment. At the same time, our basic model complex gives us several potential abilities to simulate hemodynamics also under situations when there is some hemmoragy. When we know (or can suppose) the value of blood loss, we can use this information for definition of human tolerance to the +Gz acceleration. Although this aspect was not included in the contract, we can modify and adapt our model in future if this aspect is interesting for customer.

We can see that the basic model includes 23 systemic and 6 lung arterial and venous compartments located on the different levels. Besides, the heart is presented in model by its 4 chambers. We can also see that three environments (cranial, thoracal, abdominal cavities) are special presented as extravascular environments with their specific conditions of extravascular pressures' dynamics. So, having such presentation of vascular net we are able to simulate important influences of extravascular pressure changes in these cavities on a local hemodynamics. Having aortic arch and carotid sinus compartments we will be able to describe relations between afferent nervous activity

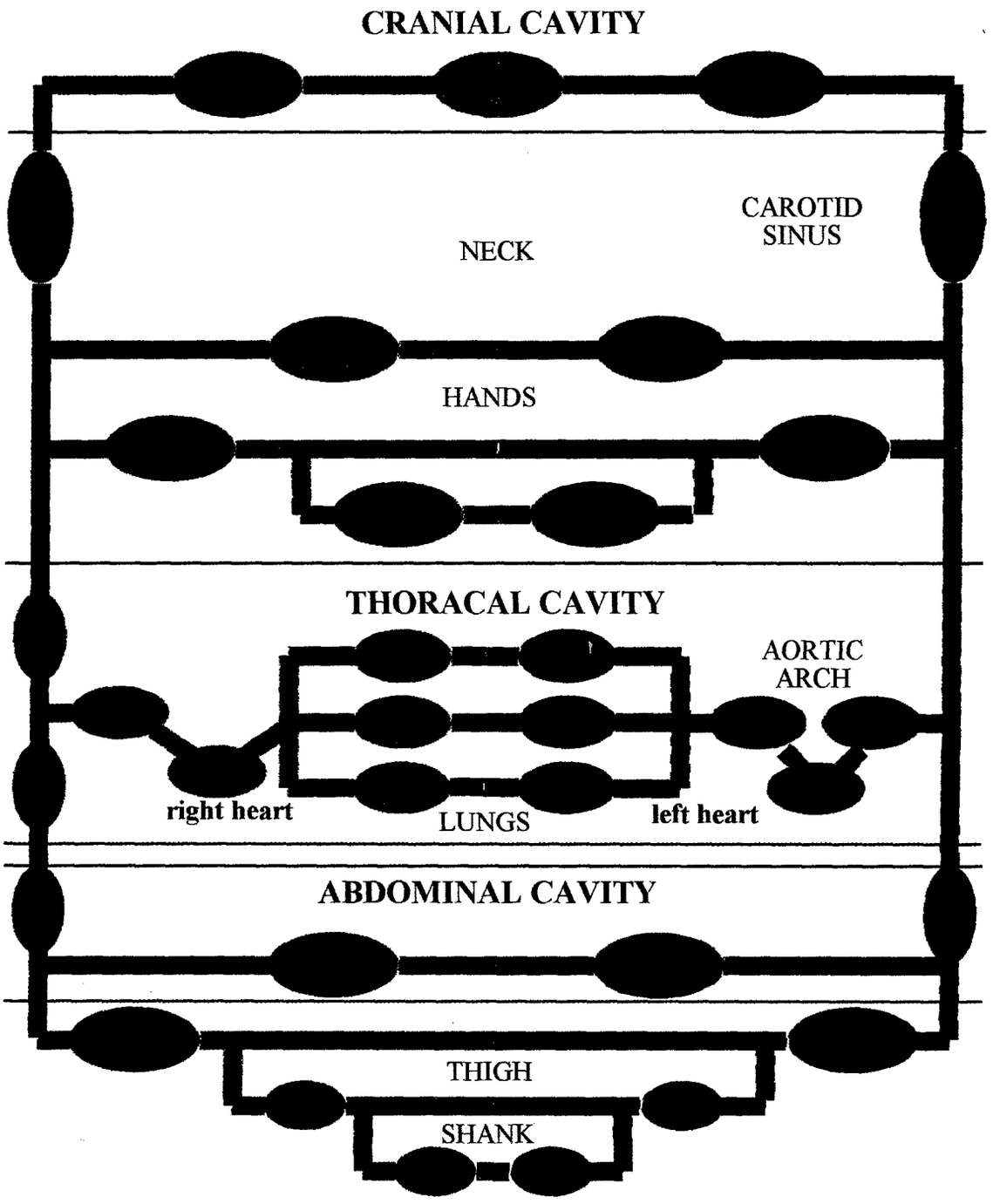


Fig. 4. Presentation of multicompartmental CVS in basic variant of model.

and transmural pressures in these reflexogenic vascular areas important for systemic arterial baroreflex .

2.3. Models of Nervous-Reflector Control of Hemodynamics

We have developed several models for description of main physiological characteristics for the central nervous-reflector control processes both under normal physiology conditions and under extremal conditions (essential change of values of gravitational factor). Generally speaking, the difference between these models was not strongly of principle for solving most of problems of applied physiology. But our experience allows us to assert that the situations with the high sustained dynamic accelerations are exclusive situations, where the hemodynamics control process can be

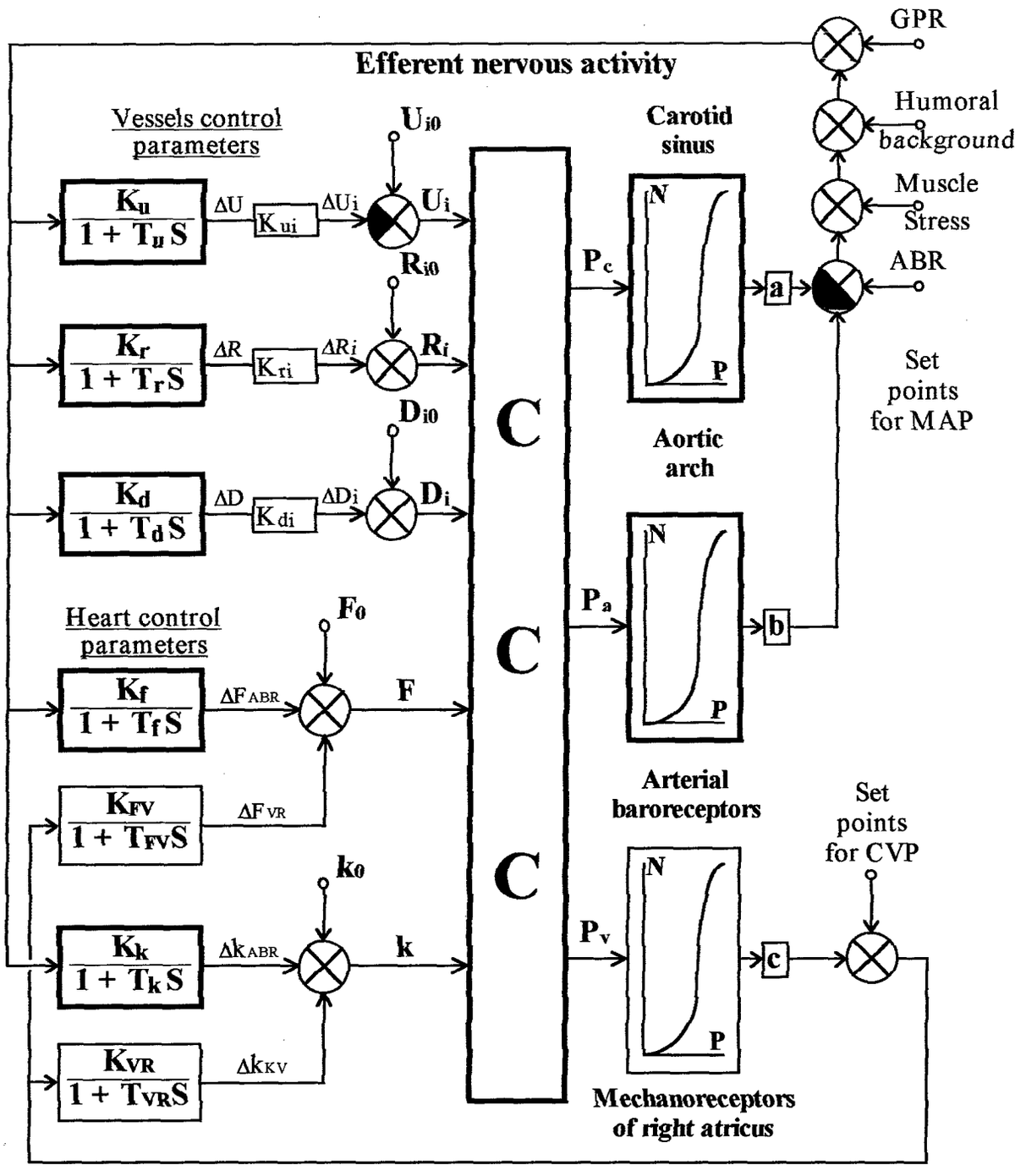


Fig.4. The general structure of model of central nervous control of hemodynamics.

provided only by nontypical combination of several self-control mechanisms of CVS. There also might be some activation of nonspecific reflexes. Such vision of inside structure of the whole central control process allows us to define and to evaluate the role and relative potentials of different mechanisms that may be involved in hemodynamic providing of human tolerance to +Gz accelerations. To analyze them it will be useful to pay attention on the fig.4, that discloses the principle scheme of cardiovascular systems reflector control under +Gz accelerations.

We can see that there are two heart control parameters (F,k) and three integral control parameters of vessels (D,U,R). The last three parameters are dispersed on different regional area of vessels, according to the existing physiological notions about their efferent sympathetic nervous density.

There are two different feedback channels on this scheme. The first one is divided in its turn and forms two negative feedback channels for arterial baroreceptor reflexes (ABR) from baroreceptors of aortic arch and carotid sinus zones. The second feedback channel is a positive feedback channel for mechanoreceptor reflex from area of right atria (this reflex is known as Bainbridge reflex). The last one can change only F and k. The ABR is often included in different models of hemodynamics control but the Bainbridge reflex is seldom presented in models. But there are sufficient reasons to include it in our hemodynamics model under piloting accelerations. The first reason is the essential increasing of central venous pressure during special breathing procedures. The second reason is that the central venous pressure increases also during muscle stress or use of extended coverage anti-G-trousers.

The nervous activity in both feedback channels is formed by the difference between set-points level for central neuronal structures and summary activity of receptors. ABR is presented in the model as proportional regulators. They are based on nonlinear S-form characteristics between arterial transmural pressures in aortic arch or carotid sinus and their summary baroreceptor activity. These characteristics take into consideration all known peculiarities of distinctions between threshold pressures and activity of baroreceptors from both zones:

$$F_j^R = \frac{1-\exp\left(\beta_j^R \cdot \left(P_j^R - P_j^t\right)\right)}{1+B_j \cdot \exp\left(\beta_j^R \cdot \left(P_j^R - P_j^t\right)\right)}$$

$$\frac{d\Delta X_i(t)}{dt} = \frac{K_i(t)\cdot E(t) - \Delta X_i(t)}{T_i}$$

$$K_i(t) = \begin{cases} K_i^P \cdot \left[X_i^{max} - X_i(t)\right] & , E(t) > 0 \\ K_i^d \cdot \left[X_i(t) - X_i^{min}\right] & , E(t) \le 0 \end{cases} \qquad (2.13)$$

$$X_i(t) = X_{oi} + M_{ij}^1 \cdot \Delta X_i(t)$$

$$E_j = P_j - p_j$$

$$X_i(t) = XB_i^{min} + \sum_{i=1}^{n} \Delta X_i(t)$$

Preliminary investigations of basic models have shown that we can not specify the characteristic behaviour of heart rate and provide the hemodynamics tolerance to +Gz more than 4,0g, if we remain only in the frame of classic notions of nervous-reflector ABR control of CVS with constant level of central set-point of mean arterial pressure.

There are some fundamental regularities of aortal and carotid baroreceptor reflexes interaction. As a rule, for most of situations that characterize intact baroreflex function, the whole reflector control is a sum of synergetic reflexes from different receptor area. The organization of general baroreflex control may sometimes have its peculiarities that may play essential role [11]. We would like to note that +Gz accelerations should be analized as a specific artificial situation with nontypical displaying of complex reflector control mechanisms [13]. Such vision of problem is important because till now the aspect of essential transformation of different baroreflexes interaction regimes remains still out of experts sight.

Aortal and carotid baroreflexes may have pressor or depressor character, depending on a level of local transmural pressures. The interaction between aortal and carotid baroreflexes may be inverted from its usual synergetic form to antagonistic one during +Gz accelerations. The antagonistic interaction takes place when the mean transmural pressure in carotid sinus becomes lower than it was under g=1.0, but the mean transmural pressure in aortic arch exceeds its level under g=1.0. Since that moment the summary effect of ABR will be less than the effect, when both reflexes were pressor. As a hemodynamic result of this situation we will see that both heart and vessels parameters begin to decrease. Such behaviour of parameters has to decrease the mean arterial pressure and thus the human tolerance to +Gz accelerations could be some lower than it is really. As we know that human +Gz tolerance is essentially more than 4,0g, we must think also about the activation of some other reflexes that can modify effects of ABR. The general pressor reflex (GPR) was included in the model as one of such possible reflexes. It is necessary to underline that GPR is only a hypothethic reflex. Moreover, this GPR-reflex is practically not mentioned in special literature. Therefore, we can only imagine mechanisms of its activity. Because the mean brain flow does not essentially decrease under these conditions, we assume that GPR may be initiated by different mechanoreceptors of overstretching structures (muscles, diaphragm, organs of thoracic and abdominal cavities).

In addition to peculiarities described above, this model also takes into consideration the changes of arterial set-point level. They may be initiated both by muscle activity and changes in level of cathecholamines in blood. For the last case we differentiate three levels of general stress (low,middle and high), indicated by value of heart rate for phone (rest) situation.

Every central reflector control mechanism was simulated as a proportional regulator that has its gain (K_i) and time constant (T_i). K_i characterizes the power, when T_i characterizes the value of inertia of reflector processes. We can simulate various conceivable hemodynamic situations by different combinations of changes in these parameters. It is necessary to note that every K_i does not have a constant value. There are some functional relations between K_i and control parameters (X_i): K_i decreases simultaneously with the increasing of parameters. Consequently, the functional reserves of CVS's control parameters have to decrease.

According to the second model of reflector control of CVS there are no set-point in central nervous control structures of brain for control of arterial pressure. All baroreflector processes may take place due to reciprocal relation between arterial baroreceptor activity, on the one hand, and efferent sympathetic and vagus neuronal activities, on the other hand. The crosspoint of these curves has to define the current level of efferent nervous activity and values of vessels and heart parameters. Appropriate equations for this variant of model look like:

$$E_s = \frac{\left(1+H_s\right)\cdot\exp\left(-H_s^1\cdot S\right)}{1+B_s^T\cdot\exp\left(-H_s^1\cdot S\right)},$$

$$E_v = \frac{b_v\cdot\exp\left(b_s\cdot S\right)}{1+d_v\cdot\exp\left(b_s\cdot S\right)}$$

$$S = \sum_{j=1}^{5} W_j\cdot F_j\left(PP_j\right)$$

$$F_j^R = \frac{1-\exp\left(\beta_j^R\cdot\left(P_j^R-P_j^t\right)\right)}{1+B_i\cdot\exp\left(\beta_i^R\cdot\left(P_i^R-P_i^t\right)\right)}$$

$$F(t) = FA + KF_s \cdot E_s(t) - KF_v \cdot El_v(t)$$

$$k(t) = kA + Kk_s \cdot E_s(t) - Kk_v \cdot E_v(t)$$

$$D_a = D_a^s \cdot E_s, \quad D_v = D_v^s \cdot E_s \qquad (2.13a)$$

$$V_p = V_s - U = \begin{cases} d_1 \cdot (a_2 \cdot E_s + b_2) & , E_s > 4 \\ d_1^u \cdot (a_3 \cdot E_s + b_3) & , E_s \le 4 \end{cases}$$

The first variant of ABR-model realizes the hemodynamics control under gradual accelerations whereas the second one is used for rapid accelerations.

In addition to the described models of CVS own reflector mechanisms, our model complex includes also some assistant reflexes that might have outside origin (relative to the CVS structures). But at the same time we think, that having some common tracks in brain structures, these assistant reflexes might influence the normal function of own reflexes of CVS included in the model and essentially modify their hemodynamic effects. According to this concept and taking into account the well known physiologic concepts, we assume that simultaneously to the increasing of general body muscle activity the value of central set-point defining the level of mean arterial pressure, must also increase. Besides, this arterial pressure's set-point has to increase also with the increasing of concentration of different humoral cardio- and vasomotor active substances (especially cathecholamines). Therefore, this aspect was also taken into account in our models. At last, the GPR mentioned above also shifts the arterial pressure set-point. Because of a lack of any quantitative data concerning these three additional mechanisms in physiology they were included in mathematical models only as a hypothetical reflexes. All quantitative coefficients were chosen by means of eurystics and tests.

$$Y_{cat} = Y_{cat}^{max} \times exp\left[-\omega \times (T - T_{exp})\right] / \left\{1 + \phi \times exp\left[-\omega \times (T - T_{exp})\right]\right\} \qquad (2.14)$$

$$Y_{gpr} = K_{gpr} \times Y \times \left\{1 - exp\left[\eta \times (W_{gpr} - A)\right]\right\} / \left\{1 + \vartheta \times \left[1 - exp\left(\left[\eta \times (W_{gpr} - A)\right]\right)\right]\right\} \qquad (2.15)$$

2.4. Modeling of technology of human protection under Gz accelerations.

All four main methods of human artificial protection under high sustained +Gz accelerations (changes of angles between human body parts and direction of acceleration vector, use of anti-G suits, muscle tense, special breathing regimes under positive pressure) were included in our final software model complex. The future user will be able to choose each of the possible combinations of protections from this list.

During the previous developments we had encountered some problems in this area. Experts in this area know well that all four methods have their limits to increase human tolerance to +Gz. At the same time nobody was able to answer concrete questions about mechanisms determining quantitative relations and limits of these protections use. In fact, all attemts to make clear these mechanisms were unsuccessful. We think, that is the main cause why the biophysically correct models of investigating processes till now aren't developed. The only way to simulate the protection technologies is to find such an euristic approximations that will allow to approach step by step to the appropriate mathematical equations that will give us acceptable results of simulation for different test situations. This way was choosen by us. Once a satisfactory result is reached the range of simulation experiments should be extended.

To describe in model the main hemodynamic effects of use anti-G-trousers we assume that the pressure transmission process from sections of trousers into the tissue around vessels may be presented in the first approximation with the help of following differential equation:

$$T \cdot \frac{dP^I(t)}{dt} = K\left(P^I(t)\right) \cdot P^E(t) - P^I(t) \quad , g \geq g_T^{th}$$
$$P^I(t) = 0 \quad , g < g_T^{th} \tag{2.16}$$

where T -time constant and K- transmission coefficient of different human body cavities or tissues, g_T^{th}-level of threshold of acceleration +Gz.

The equation system (2.16) describes also the process that characterizes the transmission of external supraathmospheric pressures (applied to the chosen body parts) into the vessels. The only difference in this case consists in different values of transmission coefficients. So, we are able to simulate the human hemodynamics also under application of negative pressures.

Although indisputable is the fact that the general body muscle tension mobilizes the venous returning of blood towards the heart and thus increases the cardiac output, the quantitative information about this mechanism is absent. Therefore, we are compelled to use euristics for modeling of hemodymamic effects of muscle stress under +Gz accelerations. The formula reflecting the correspondence between the mean muscle tense (P_m) and the value of accelerations starting from some threshold, looks as following:

$$P_m == \begin{cases} 0, & g \leq g_m^{th} \\ A \cdot P_m^{max} \cdot \left(g - g_m^{th}\right), & g_m^{th} < g \leq P^{max} \\ P_m^{max}, & g > P^{max} \end{cases} \tag{2.17}$$

For modeling of hemodynamic effects of breathing under positive pressure we assume that the intrapleural pressure Pp linearly increases beginning from some +Gz-threshold level:

$$P_p = P_{p_o} - \alpha_{BP} \cdot \left(g - g_{BP}^{th}\right) \tag{2.18}$$

So, the mathematical equations system numbered as (2.1 - 2.18) is the basic peoblem-oriented model complex able to describe practically all situations which may take place during development and testing systems for evaluation of human tolerance to +Gz accelerations. Before to transform this model to special computer software technology, we have discussed with experts the main demands to the software interface.

The special and important aspect of modeling is the determining of values for model's constant parameters. This stage of computer models develompent has required essential time and efforts from us. Only after several thousand computer experiments for different acceleration regimes with different combinations of and protections we have became sure that the whole software develompent is successful in general.

Chapter 3. DESCRIPTION OF MAIN ALGORITHMS FOR COMPUTER REALIZATION OF MODEL COMPLEX

According to the general conception of use of the model complex , our special software includes several basic research algorithms. Generally speaking, the main aim of all these algorithms is providing of physiologist-researchers with computer simulation of a wide range of situations that may take place during testing of human tolerance to +Gz accelerations. Besides, these algorithms must allow help to create a wide range of hypothetic situations. These hypothetic situations are related to:

1) the choosing of acceleration profiles;
2) the combinations of existing protections;
3) the prospective methods of protections;
4) several diseases.

The last aspect is subdivided into two branches. The first one is determined by the customer's list of diseases. The second branch relates to the possibility create of special different regimes in human physiologic state by means of direct access to the parameters of basic mathematical models. In case of further collaboration we intend to make explicit such kind of access for several parameters, related to the infrigments of central nervous control mechanisms.

So, according to the above-mentioned aspects of use of special software, there are two main algorithms, each with some secondary peculiarities. The first main algorithm is tuned to the computer simulation of hemodynamics of healthy men under +Gz accelerations. The second main algorithm provides physiologist-researchers in the case when they are interested in specific problems, that taking place during simulation the behavior of sick human's hemodynamics under +Gz accelerations.

The general structure of the software consists of the following main blocks:

- The block of all mathematical calculations. It is able to realize all necessary calculation algorithms according to the mathematical models included in MMC and strongly does not have any access by users after final tuning of models.
- The user interface. This software block provides users during realization of all regimes of computer simulation experiments, saving of results of simulation experiments in table, graphic and as a special hard copy forms, extraction of chosen data and also comparing different simulation experiments results.

For a particular class of users interested in regimes that were not previously included in the standard user interface we are planing further to create a limited direct access to models important parameters after users additional training. The additional block for initialization of constants of basic models by means of standard user interface could be used to construct an individual (male, female) model.

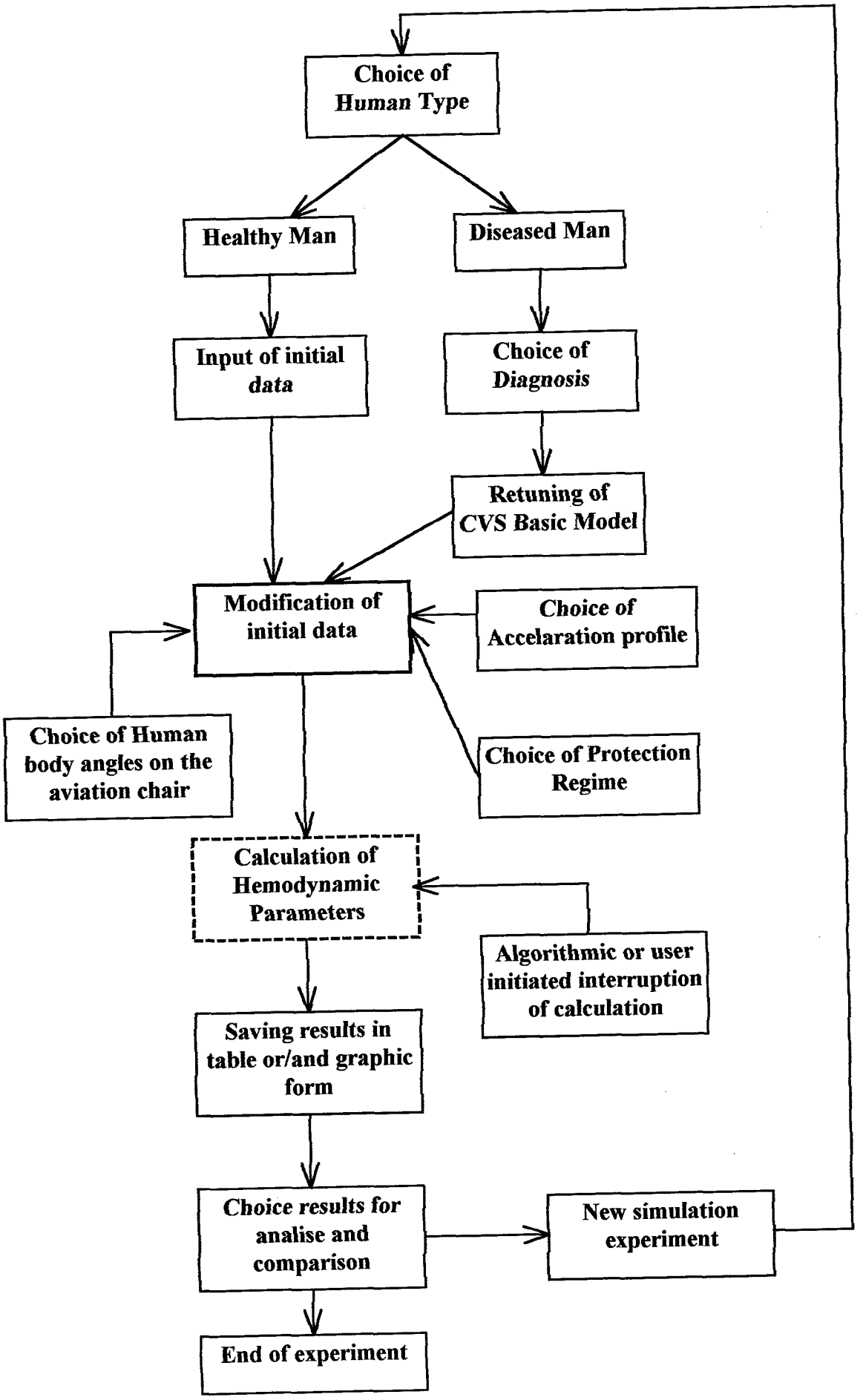


Fig. 5. The General Algorithm of Typical Simulation experiment

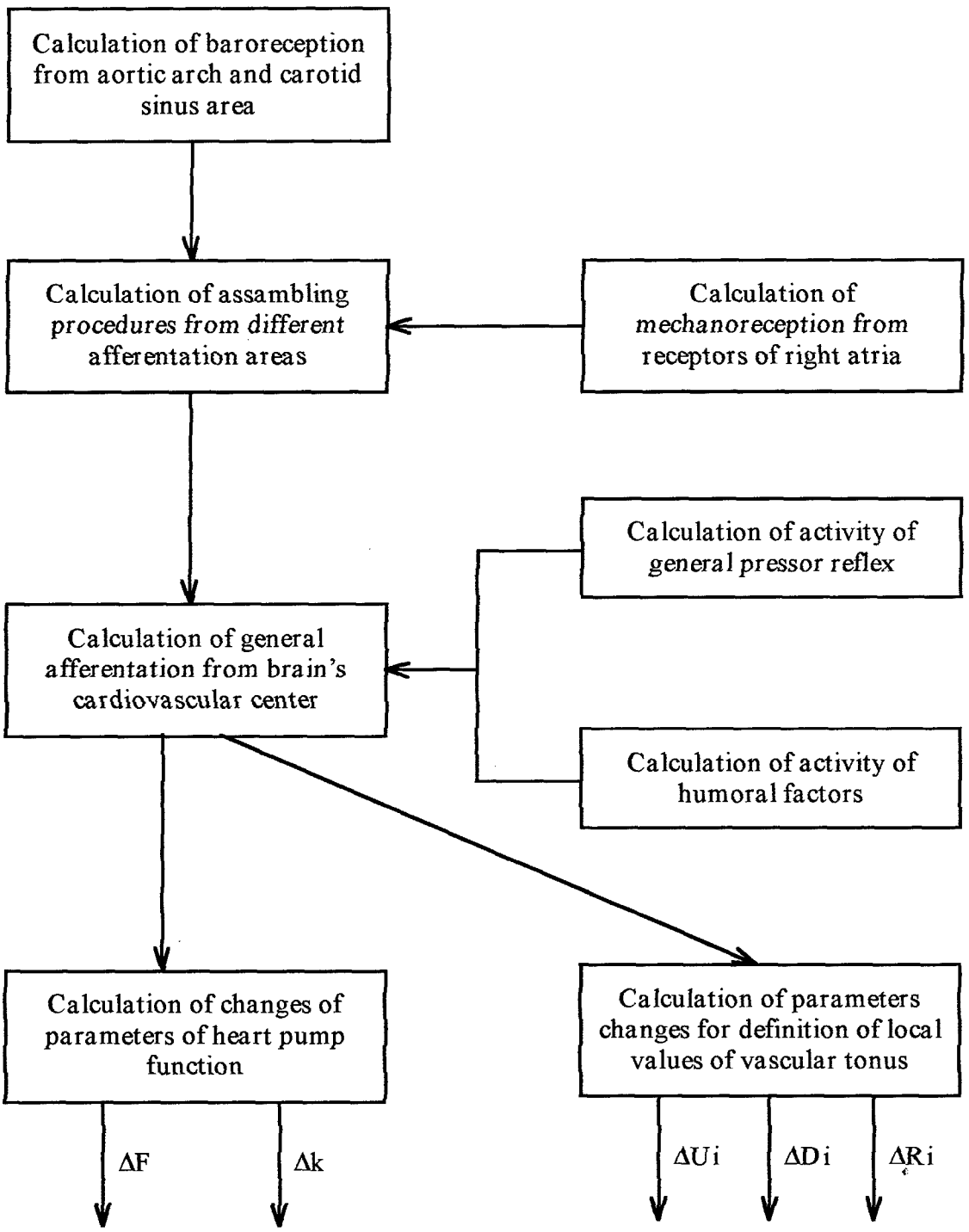


Fig. 6. Block of central control mechanisms

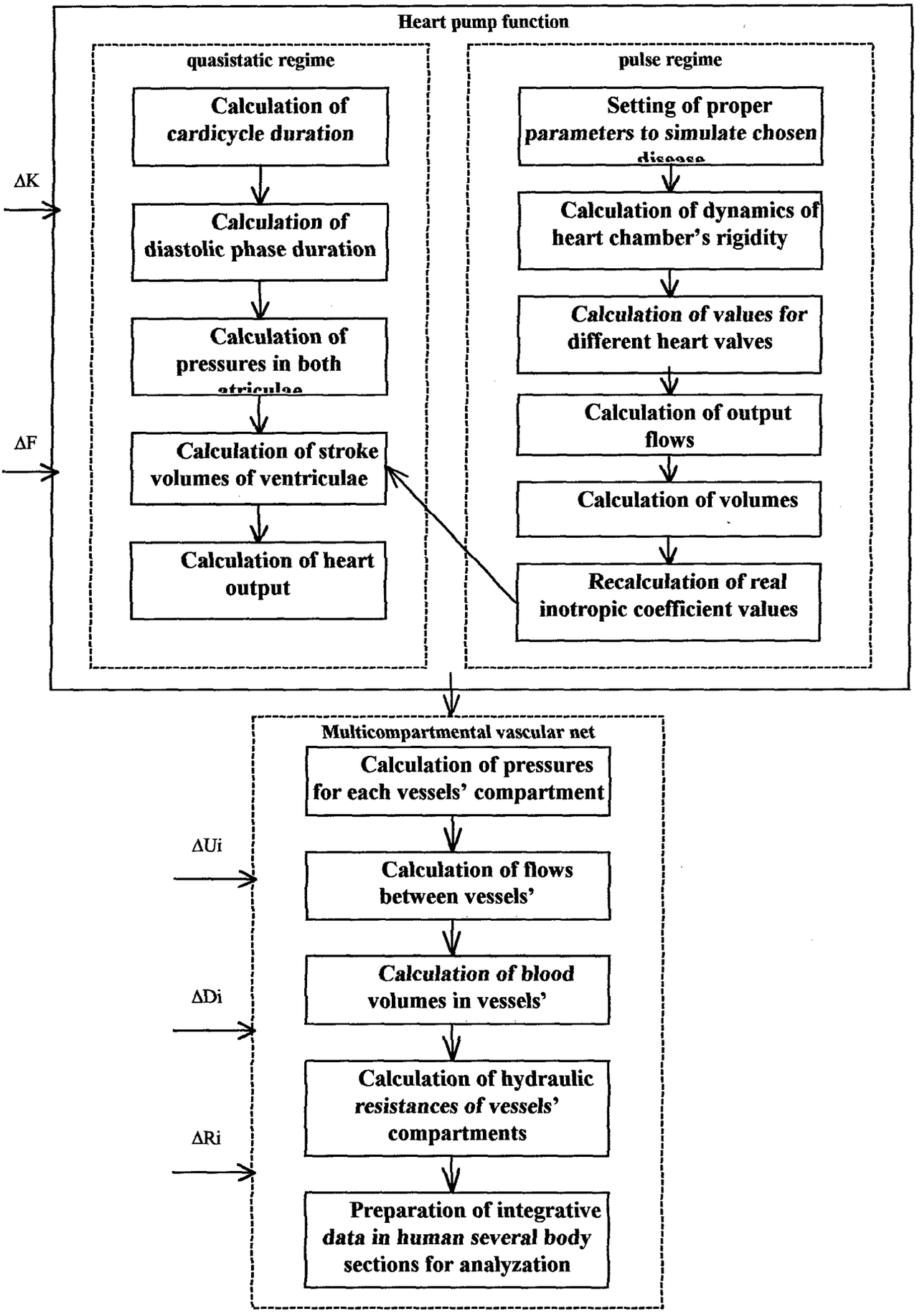


Fig. 7. The basic algorithm for calculation of parameters of human hemodynamics under +Gz accelaration

Fig.5 illustrates the main algorithm of typical computer simulation experiment. The detailed description of certain blocks of the algorithm is presented on figures 6 and 7.

There are different variants of calculation algorithms for approximate solution of system of mathematical algebraic-differential equations. The simple calculation algorithm, based on Gauss-method, has been chosen because it allows to minimize the calculation time. This algorithm has advantages and disadvantages. The known disadvantage of this algorithm is its low stability to the data values variation for starting point. First we tried to start the calculation process with a combination of models' constants that are able to provide a minimum of starting calculation time necessary to reach a steady-state regime of hemodynamics for healthy man in his horizontal position. Then the software automatically provides the changing of angles of human body parts. In fact it accomplishes the transition from clinostatica to the human sitting position with given combination of angles between body different parts and the vector of acceleration. A new steady-state regime typical for human sitting position, can be reached after a short time transitional process. Such kind of preliminary calculations were needed to avoid a possible nonstability of hemodynamics that might take place in case the simulation process starts immediately after setting of data for human different sitting positions. Although the chosen algorithm compels us to waste a little time, its advantage is that we can guarantee fluent and stable calculations. Therefore, we have kept this starting calculation algorithm in the final software. Otherwise, we would be compelled to use much more complicated calculation algorithms to avoid different possible calculation oscillations or even calculations breaking. But these algorithms would require essentially more (about six times) general calculation time. So, we think the first calculation algorithm is optimal. We are able to hide all this calculations preparations from users in case there will be such proposition. At the same time we think that physiologist can extract a useful information about stress of the hemodynamics control mechanisms in human sitting position by comparing it with the clinostatic regime of hemodynamics. This information is important to evaluate the tolerance reserves of CVS to +Gz accelerations.

It is necessary to discuss our approach to the problem how to simulate pathologic situations required by customer and presented as list of several human diseases. Analysing some of these pathologic situations we can see that the majority of them are related to the cardiopathology. Therefore, their final hemodynamic manifestation consists in the changing (decreasing) of heart pump function. As it follows from equations system presented in chapter 2 and numbered as (2.1) (see page 10) the decreasing of HPF may be simulated by means of decreasing of heart inotropic coefficient k. Therefore, we have found out the correspondence between each pathology (if we are able to simulate the gradual transition from normal physiologic regime to a disease) and appropriate value of k. In fact our current conception of diseases simulation is based on this approach.

Unfortunatelly, we are not able to proof the effectiveness of the chosen concept and appropriate algorithm at this stage of the development. We hope that we could evaluate this effectiveness after the software would be tested by users tested in special simulation experiments.

Chapter 4. SOFTWARE: DESCRIPTION OF USER INTERFACE

The main goal of software "PILACCEL", developed and presented as a special informational technology (SIT), is to give experimentators-physiologists an ability to investigate some problems of human tolerance to the accelerations Gz by theoretic way. The presented user interface (UI) is a special software which assists experimentator in his/her interaction with computer in "PILACCEL" environment.

The UI consists of four main and some additional screen forms.

The first screen form looks like an aircraft, which is the logotype of "PILACCEL" (see fig.8).

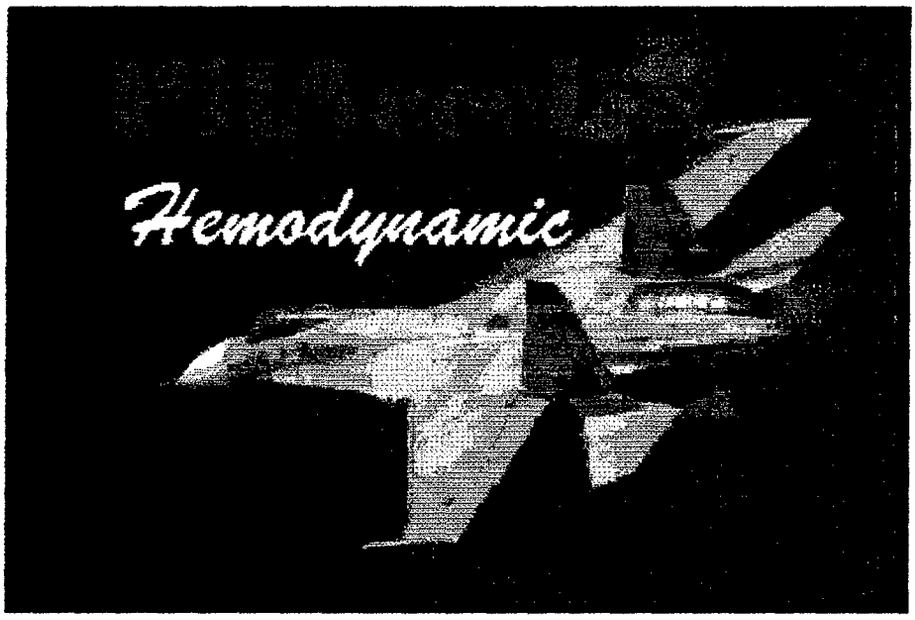


Fig. 8. The first screen form of "PILACCEL".

The second (main) screen form appearing every time user clicks on the first form field, is the main form of UI named "PilAccel - Mean"(see fig. 9).

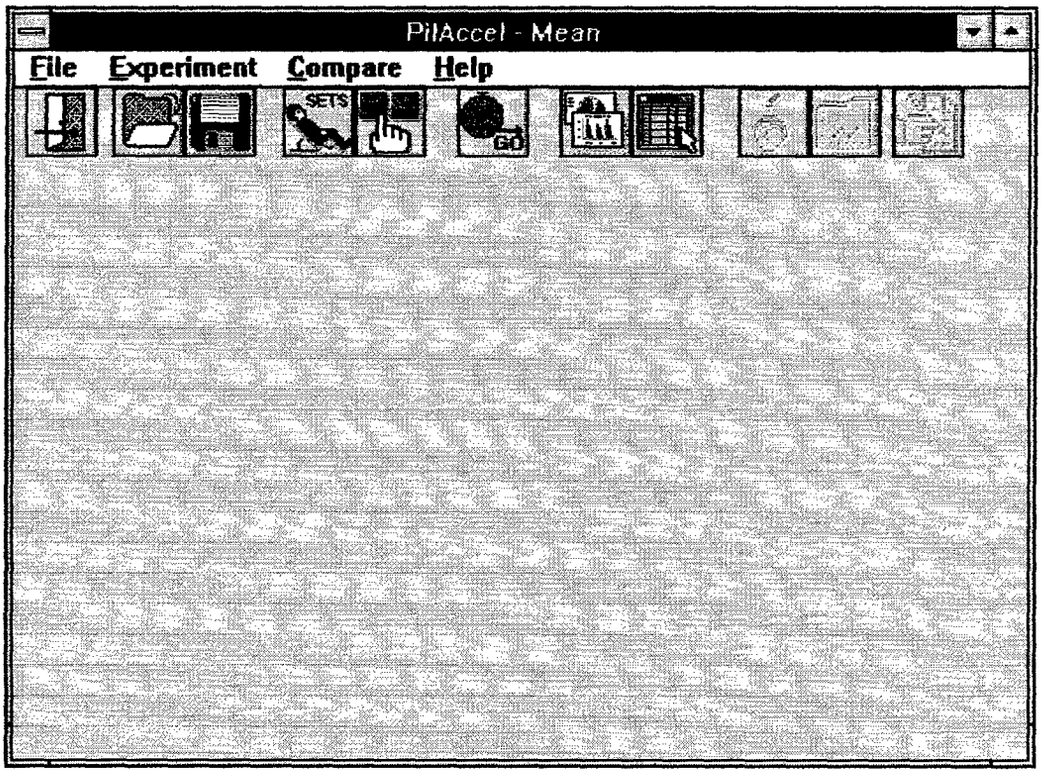


Fig.9. The main screen form of user interface

It presents visual information about "PILACCEL" structure. As you can see on fig. 9, this form of UI has an usual system of user menu, and additional 11 speed buttons. In addition to the intuitive pictograms, each of these buttons has its bottom located short informational help characteristics about function of chosen button. The button "GO" is staying in the middle part of the screen. It starts the calculation process. But user should remember that before starting calculations, user must set several experiment parameters.

The access to the setting of the neseccary for simulation experiment parameters is provided by standard user menu. The second way to do setting is to use of twin-buttons located just on the left of "GO".

When user clicks the button with the yellow subtitle "Experiment parameters", he/she is able to open the third main screen form which is the first main screen window form for settings named "Experiment parameters setting" (see fig.10).

As you can see, there are three titles: "Models", "Acceleration profiles"and "Protections". It is important to note that only one of them can be currently activated (in dash and dash line frame).

Clicking the field "Models", you open the next screen form (see fig.10).

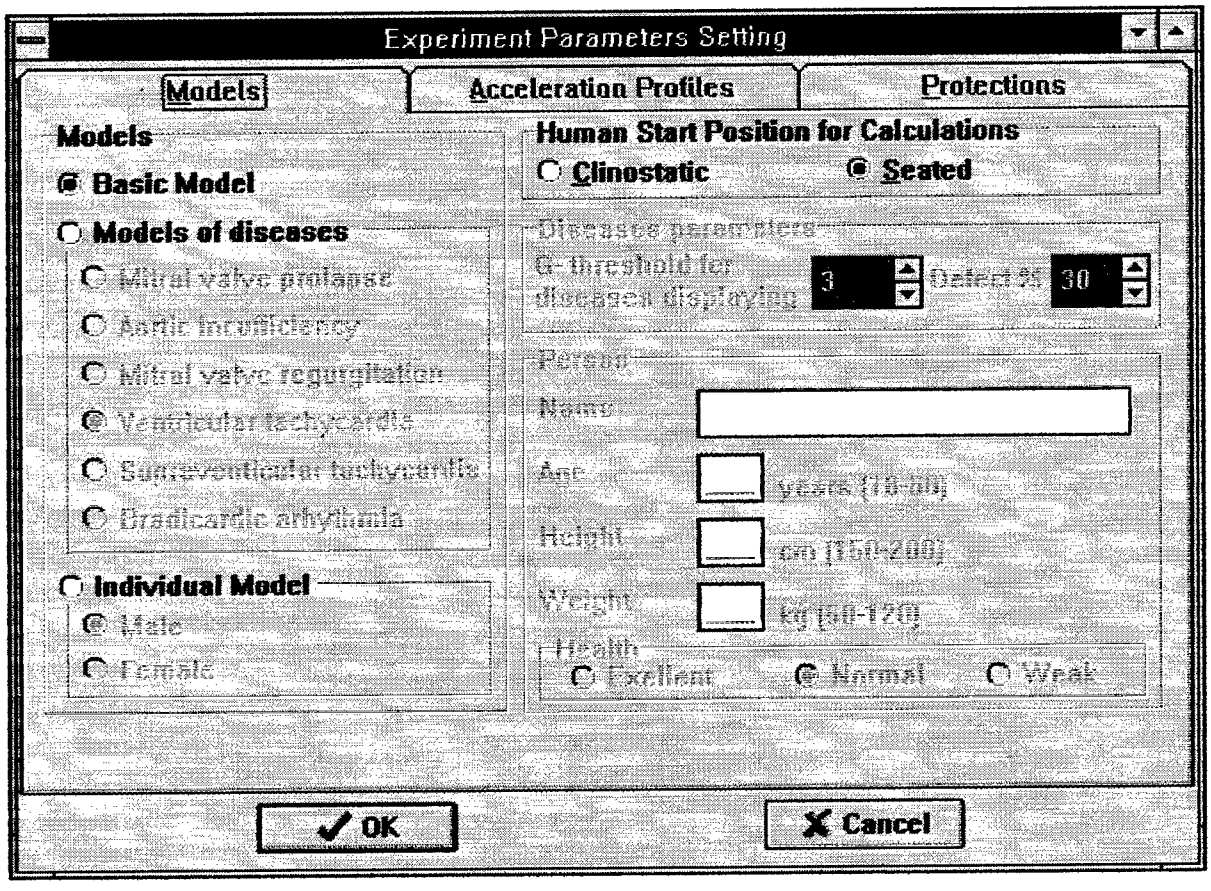


Fig.10. The main screen window form "Experiment parameters", allowing to start settings of models parameters to prepare a computer experiment. Now the field "Models" became active.

The activated field is of black color while the other fields are in gray color. You can see several model variants. But it should be noted that not all variants are accessible now. The majority part of variants are already full or partially realized while the others are displayed to show potentials and several directions for future developments. The presented variant of "PILACCEL" is able to provide reliable function only for the basic model (for mean healthy man in his seated position). We have also presented our attempts in the direction of diseases simulation, although, as it has been noted above, we are not sure about models adequacy.

Clicking the next top field of the third main screen form "Acceleration profiles" you will open a new special screen form for settings of parameters of loading acceleration regimes (see fig.11).

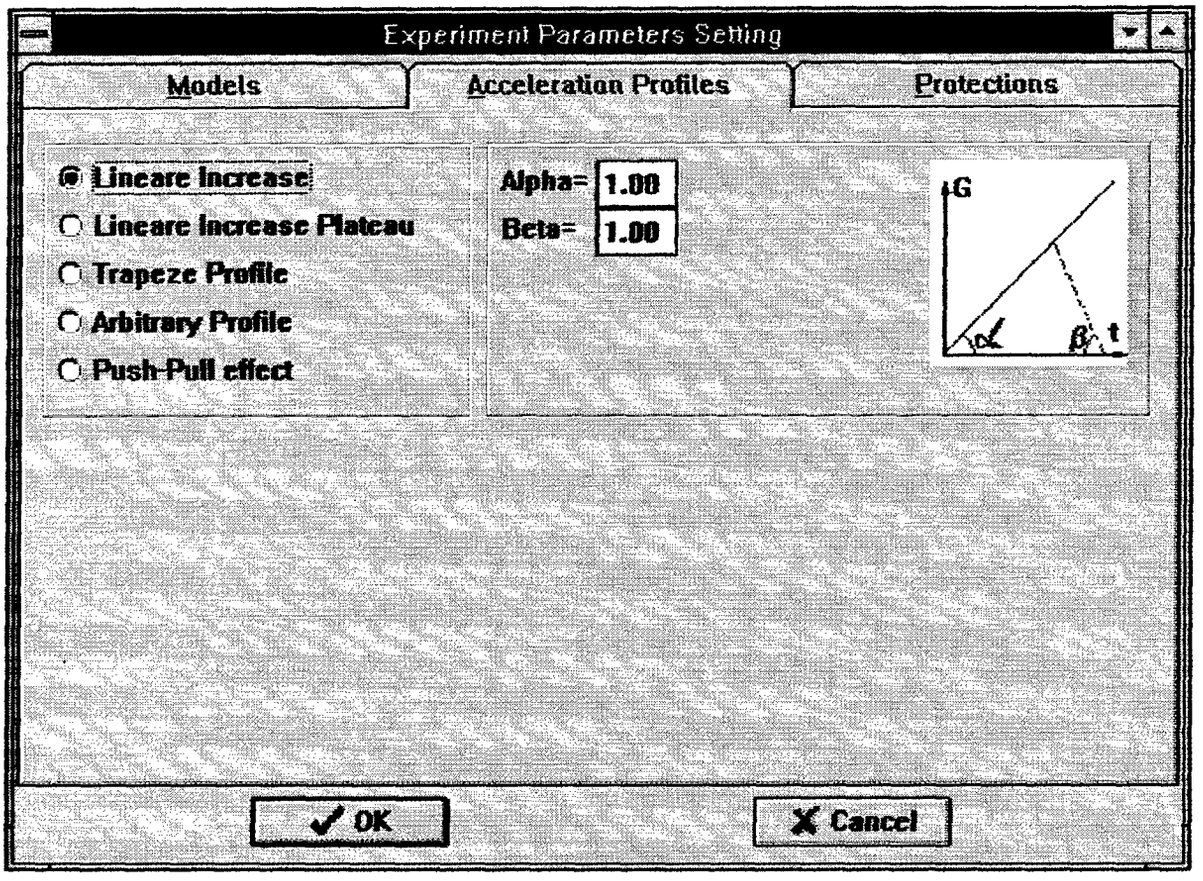


Fig.11.The general screen form for choosing of acceleration profiles. The linear regime for acceleration increasing with appropriate values of setting parameters is chosen.

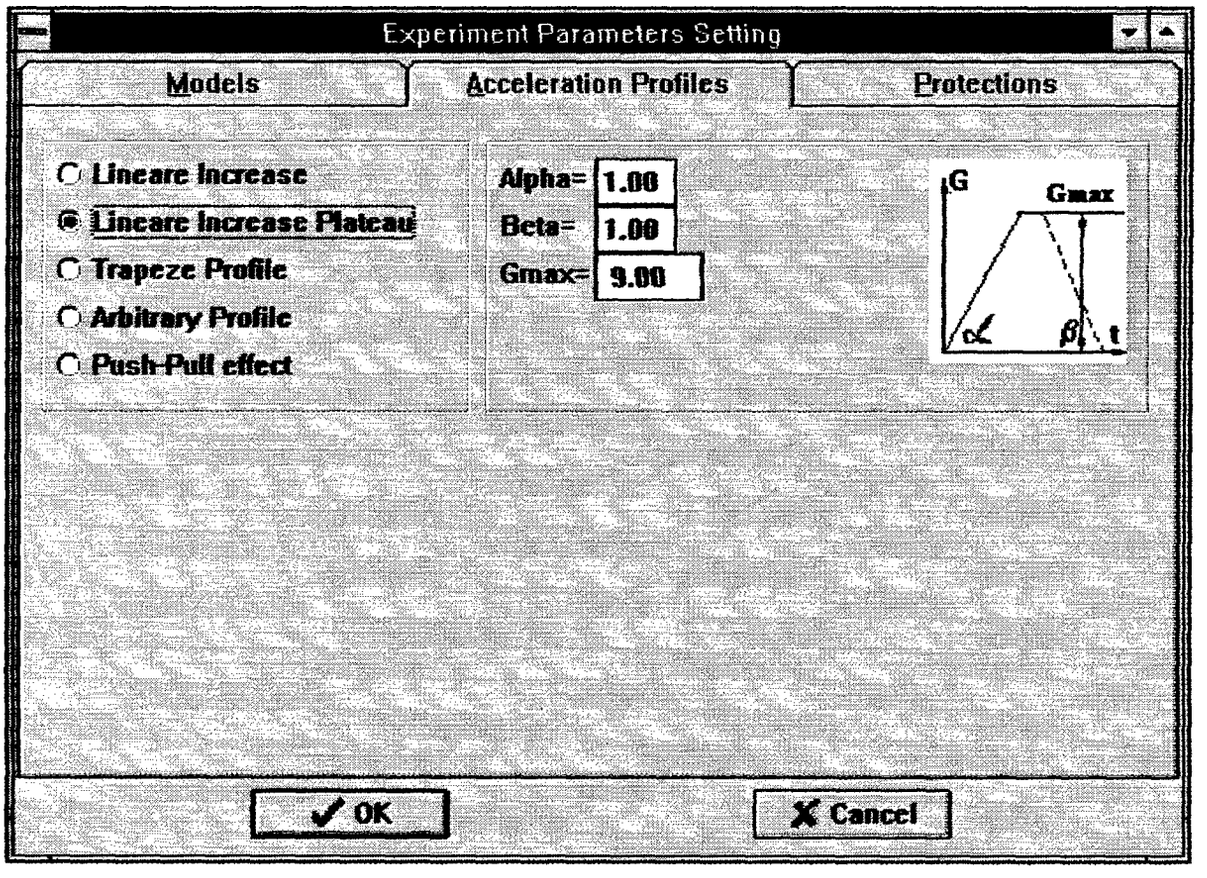


Fig.12.The standard screen form to set an acceleration profile with linear increase up to a plateau.

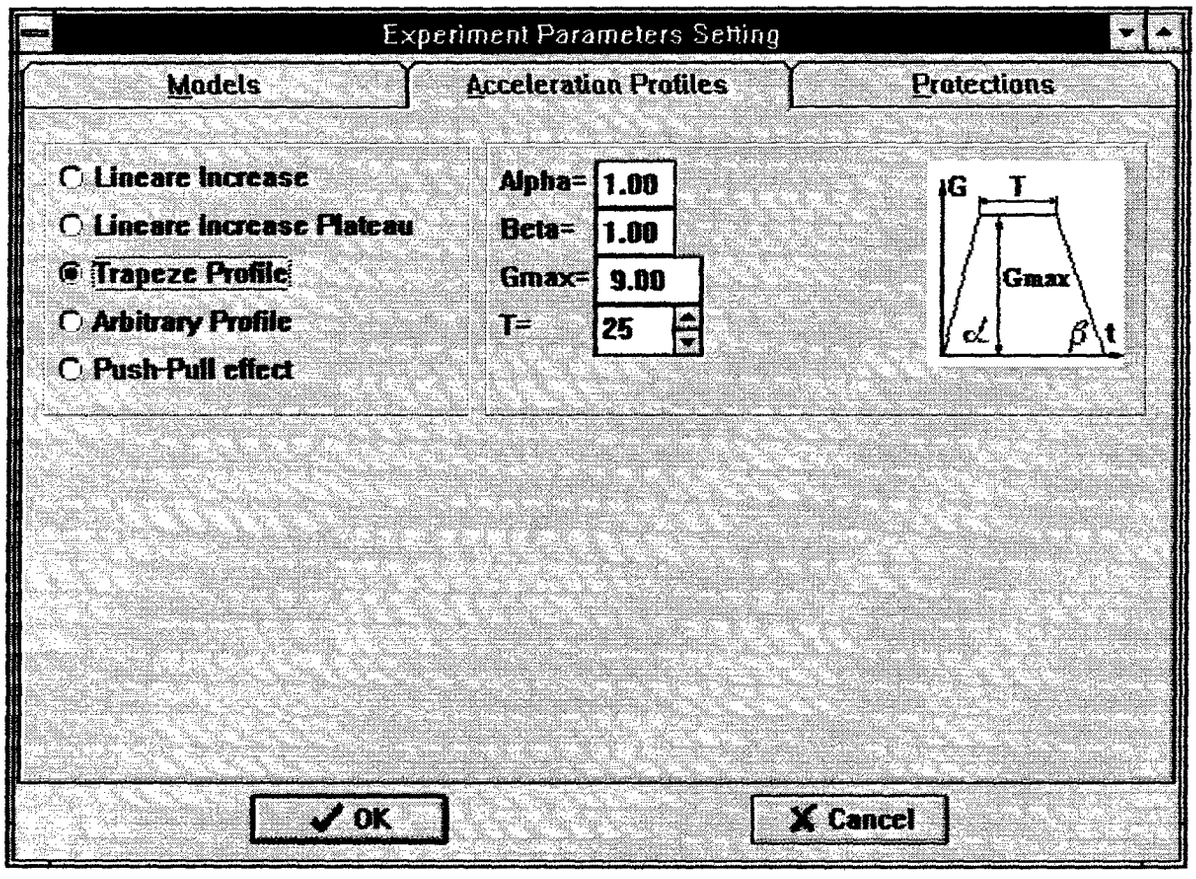


Fig.13.The standard screen form to set trapezoid form acceleration profile.

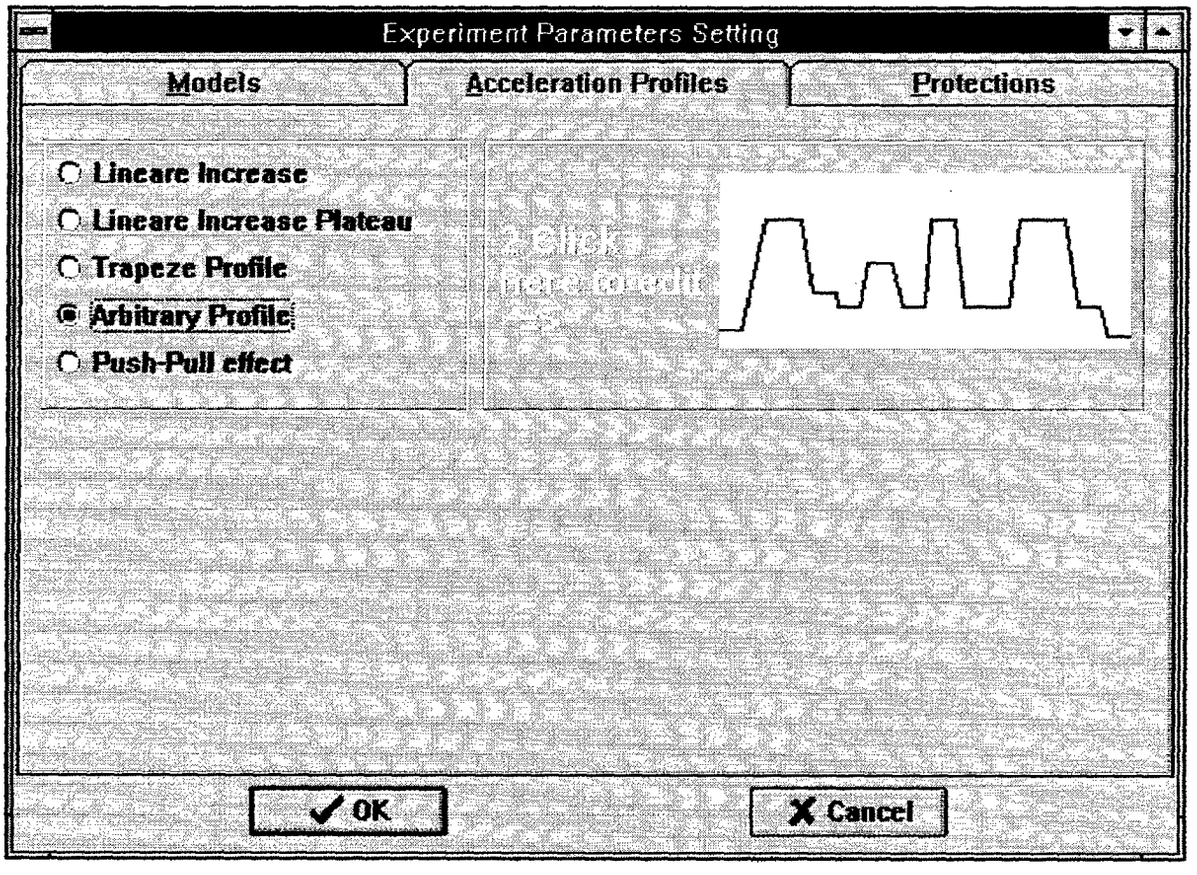


Fig. 14A. The main screen form to set an arbitrary acceleration profile.

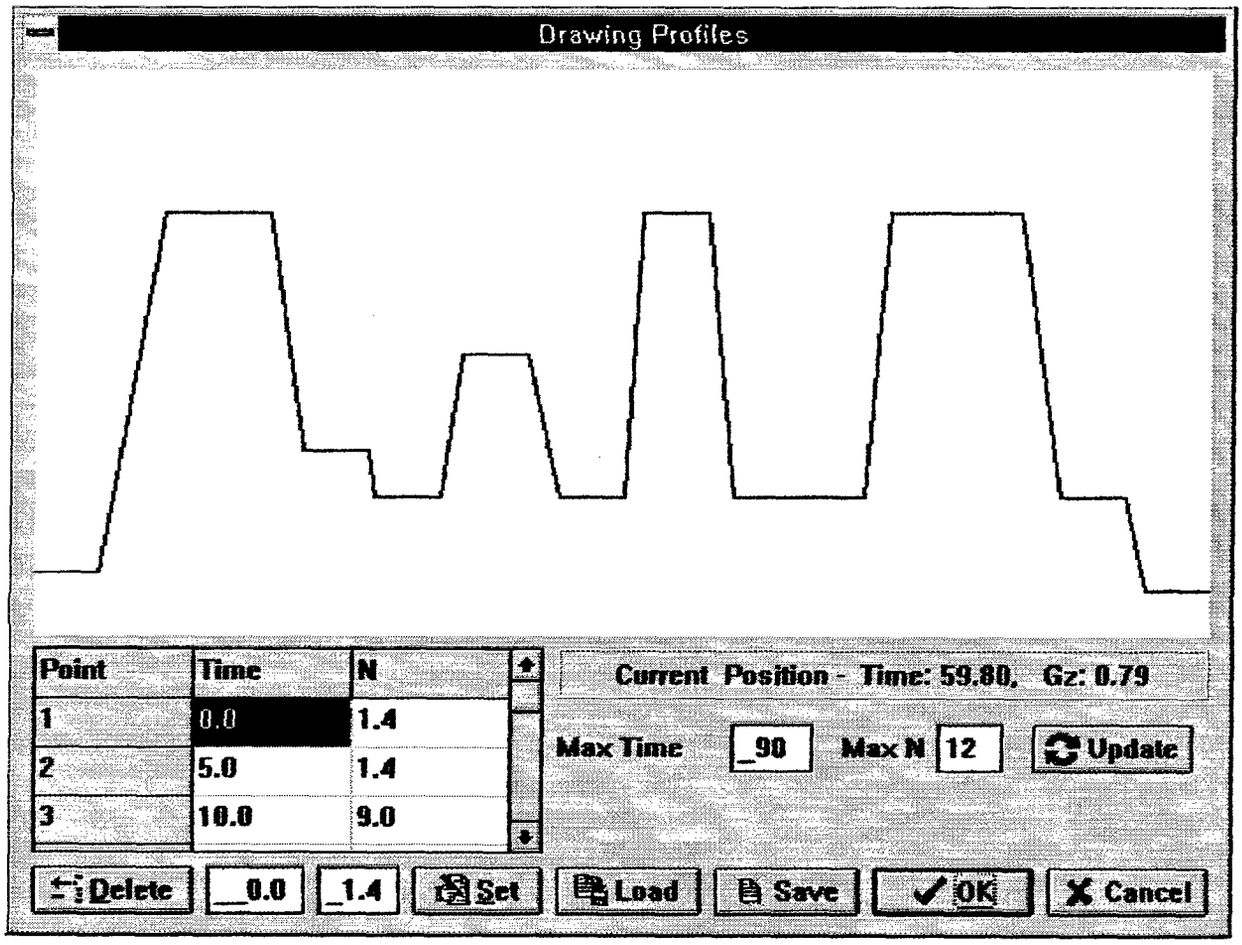


Fig.14B The assistance screen form to set all parameters of arbitrary profile.

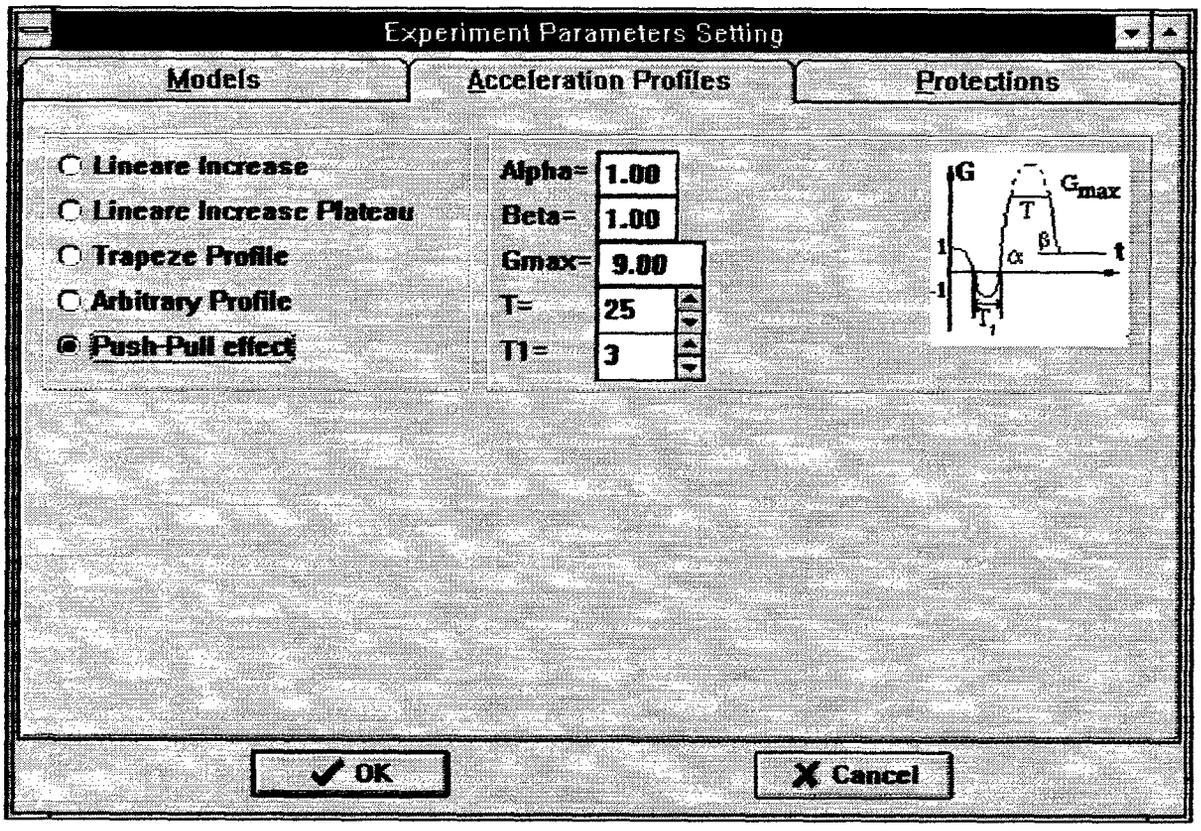


Fig.15.The screen form to set the experiment parameters for simulation of acceleration profile

Clicking twice on the special white color right field of the main screen form illustrated on fig.14A, the user gains access to the special window of assistance screen form to form a new arbitrary acceleration profile. The chosen profile might be served and later quickly loaded if is necessary.

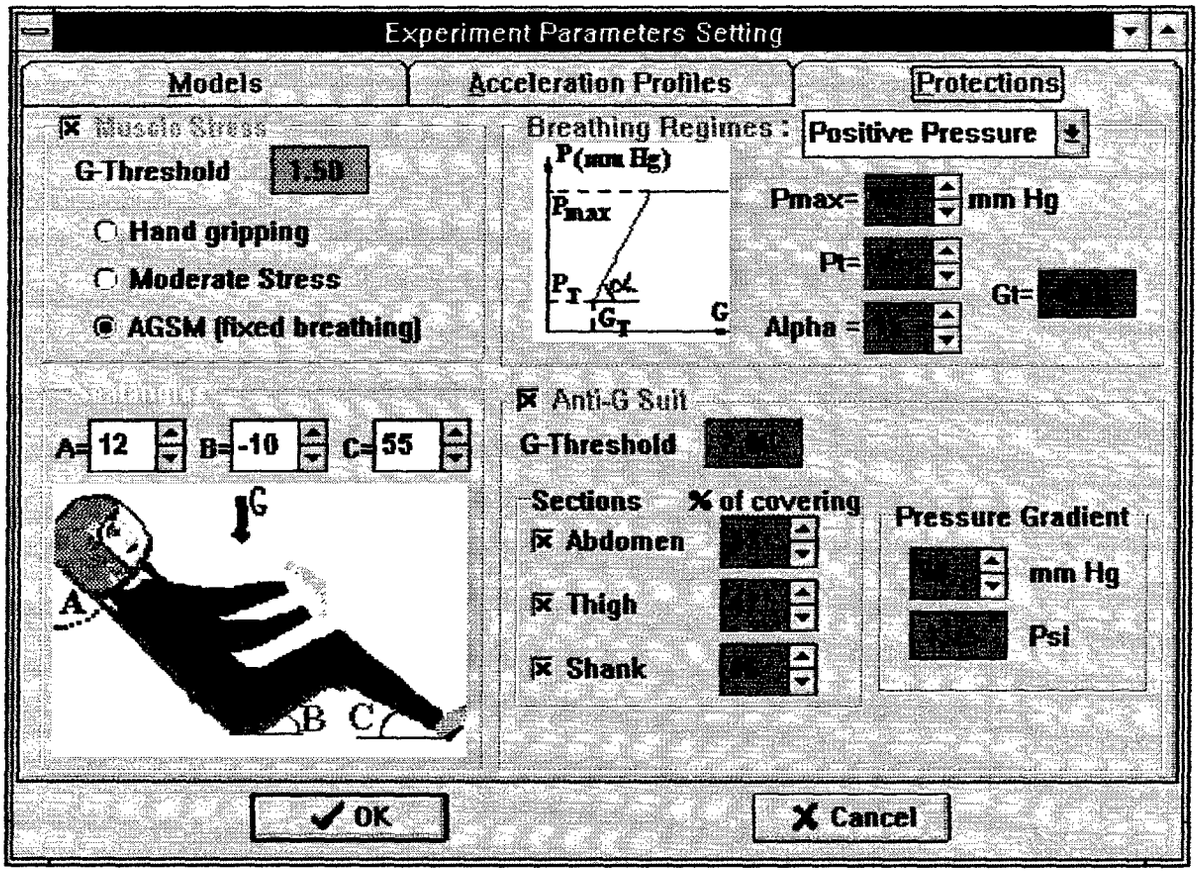


Fig.16. The screen form to set actual parameters of models of human protection.

Clicking "Protections" you can open a new window with four protection groups (see fig.16.):

- Seat-angles;
- Muscle state (relaxed when it is gray). The available variant for the AGSM simulation is fixed. It means that you cannot change the time intervals between expiration and inspiration. They are fixed on the values 1sec. and 4sec.

Besides, you can see the breathing originated pulsation of parameters only after begining of Gz loading (Although the model is able to show these pulsations also for human phone seated position before G-onset.);

- Breathing regimes (only two variants are available now - the natural breathing and the breathing under the positive pressure with possibility of its parameters setting);
- Anti-G suit (you can also set different parameters).

You should remember: it is not the all yet!

You should also turn your attention to the second button of above-mentioned twin-button.

Clicking the right button called "Hypothesses", you can do some of settings concerned central nervous control of hemodynamic situation just before Gz onset. In general, you can investigate the role of central nervous mechanisms in providing the human tolerance to the Gz accelerations, including the special situation when this control is absent.

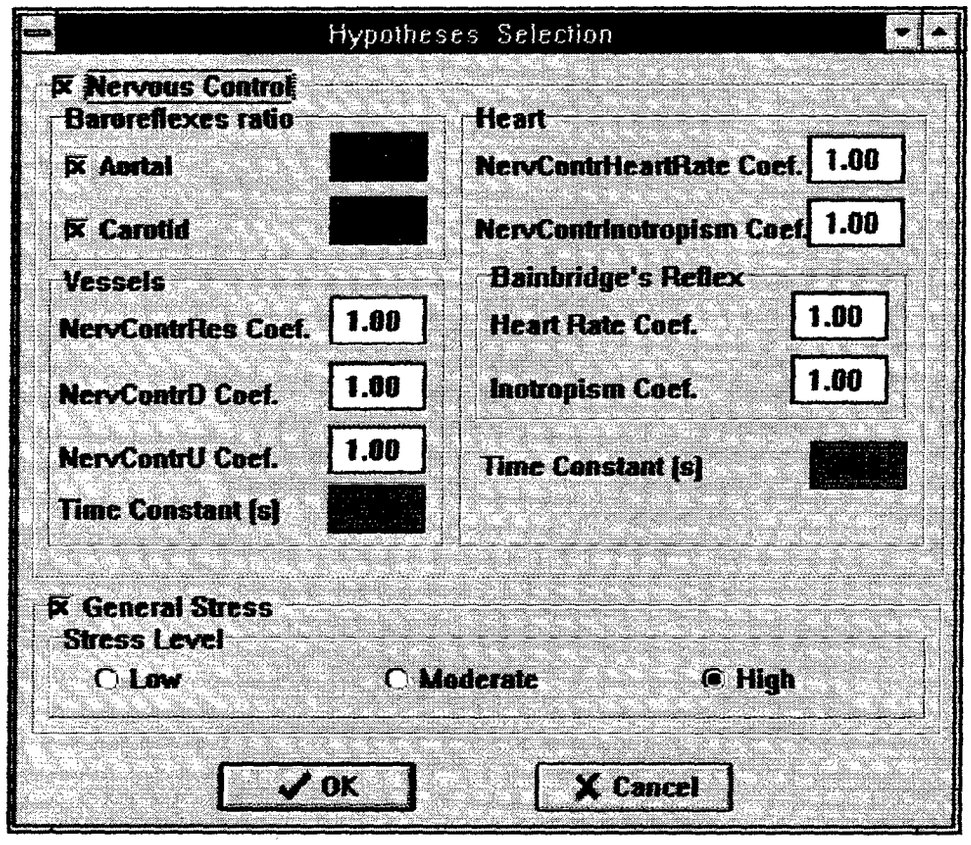


Fig.17. The screen form to set actual parameters of models of central nervous control of CVS.

It is possible to obtain different combinations of the proposed mechanisms activating or deactivating the appropriate central nervous control mechanism or their several CVS-parameters. Besides, you can set also different levels of coefficients and values of time constants of heart and/or vessels control parameters. In this case you can use the following parameters of nervous activity :

- heart rate (F) and heart inotropism (k)- to control of heart pump function ;
- the arterial resistance (R), the venous volumal rigidity (D) and vessels' unstressed volume (U)- to control of vascular tonus parameters.

If the value of a coefficient is 1, it means that we have a regime of physiological norm (excluding the coefficients for aortal and carotid baroreflexes, that normally equal 0.5). Below located mini help-windows inform user about limits of coefficients changes.

This window also allows to set the level of nervous stress of a person just before Gz onset (Assistance information about heart rate value for each of three variants of stress level is also presented in the mini help-windows located below).

Such combination of settings is named "Experiment Option".

After all these settings are performed, you can press "GO" and start the computer simulation experiment. Information confirming the start will appear on the screen in the form of special window "Calculation in progress". This window contains a running circle on its left side. The right field informs about running time, running and maximal Gz levels. When you start for the first time and also every time after you have changed some of seat-angles, you should first wait a fixed time (30s) untill a new hemodynamic stable regime for starting (phone) position is reached. The appropriate information "Hemodynamics has reached its steady-state regime for G onset" will appear on the screen.The calculation, visually indicated also by running circle, will stop. Then you should press "OK" and go on.

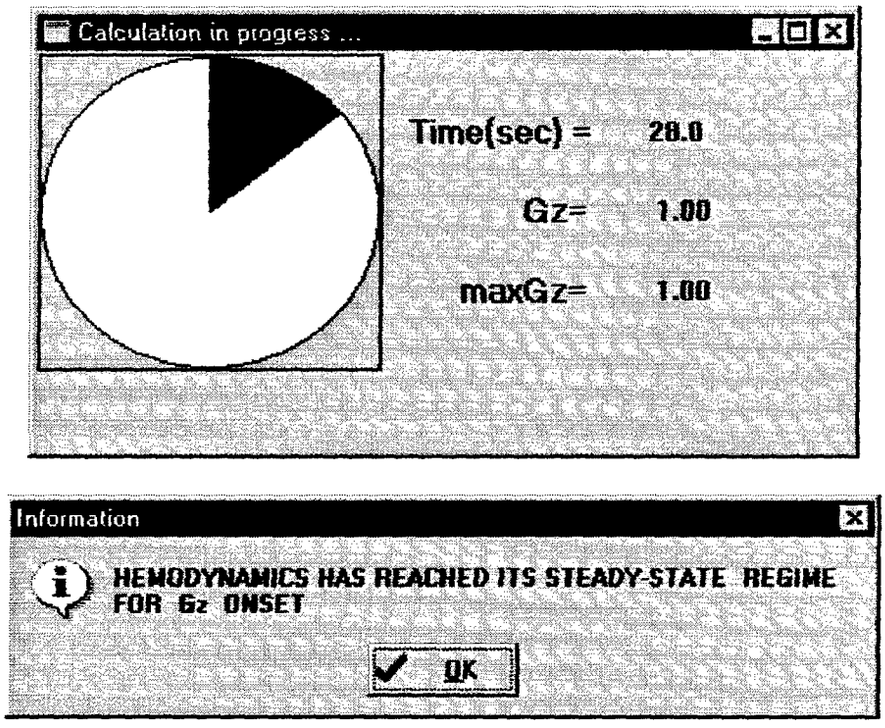


Fig.18. Screen forms after the calculation has started.

Calculations will continue until all the planned acceleration profile is completed or the sistolic arterial pressure in eyes falls down below 30 mm Hg. This situation is defined as a CLL (central light loss). When CLL takes place, "PILACCEL" informs about it by warning appearing on the screen form presented on fig.19. Having received this warning the user should use one of three buttons. If you choose "Cancel", you interrupt the acceleration's increasing and you are transferred to the regime of decceleration. In this case the recovering processes in hemodynamics starts. So, the dropped arterial pressure begins to increase again. When it becomes higher than 30 mm Hg, "PILACCEL" informed with the next screen form (see fig.20).

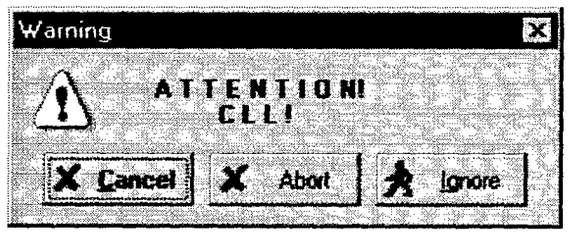


Fig.19. The screen form that informing about stopping of calculations because of CLL.

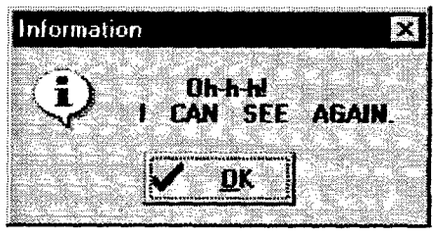


Fig.20. Screen form for indicating the end of CLL .

If you choose the button "Abort", you break all calculations and the "PILACCEL" returns again to the starting regime.

The scenario will differ when you have pressed the button "Ignore". It means you will continue increasing of accelerations level in spite of the warning about CLL. In this case some time later the blockade of brain perfusia might occur because the arterial pressure continues to decrease. The appropriate information will appear as it illustrated on fig.21.

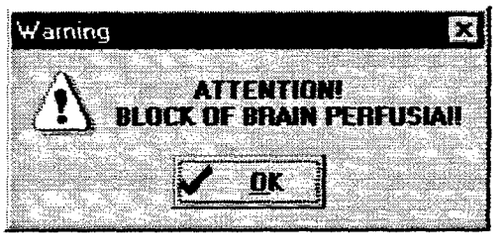


Fig. 21. Screen form warning the blockade of brain perfusia.

According to the biochemical mechanisms, approximately 2sec after the brain perfusia is blocked the human consciousness will be lost (LOC). If this situation happens you must break the loading process and start the deccelleration.

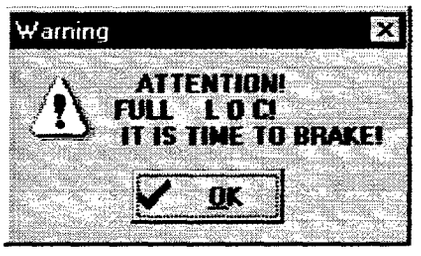


Fig.22. Screen form warning the loss of consciousness (LOC).

For all of three chosen regimes the "PILACCEL" fixes parameters for top level of accelerations (see fig.23). These parameters are also indicated on the special graphic forms of results presentation of results (you can see it as a vertical red line on appropriate figures in the next chapter.).

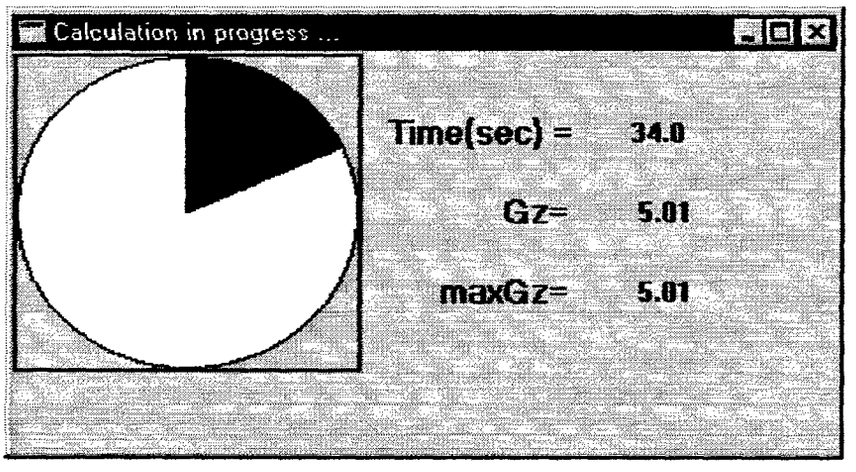


Fig.23. Screen form to fix time and value of maximal level of Gz accelerations.

The fig.24 illustrates that Gz and maxGz levels are different after the vision is recovered.

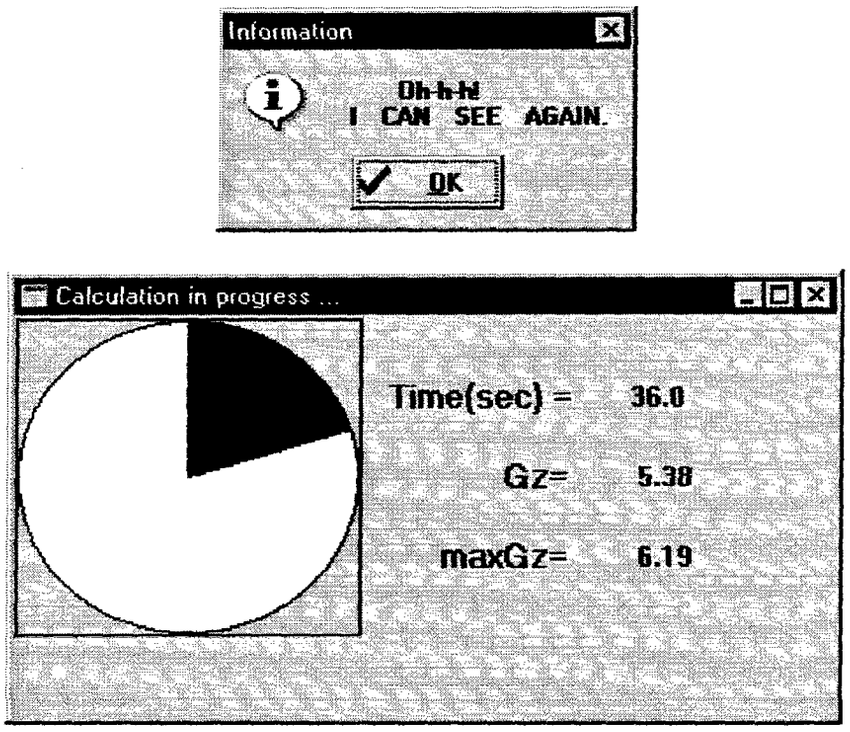


Fig.24. The current value of Gz after decceleration is less than its maximum when CLL or LOC was fixed.

How to use Results of Simulation Experiment?

The software "PILACCEL" provides users with three forms for watching and analyzing the Results of Simulation Experiment: Graphic, Table and Special Experiment Protocol. The appropriate speed buttons are located on the right of the button "GO". These buttons can not be activated at the start of the experiment (they have green color). They become accessible just after calculations are done.

If you want to compare results of two simulation experiments, you should first save the results of current experiment using the button located just on the right of "Table". Then, conducting the next experiment, you can use the button "Compare". It is possible compare only two graphics of one variable. The list of these variables is available (see fig.25).

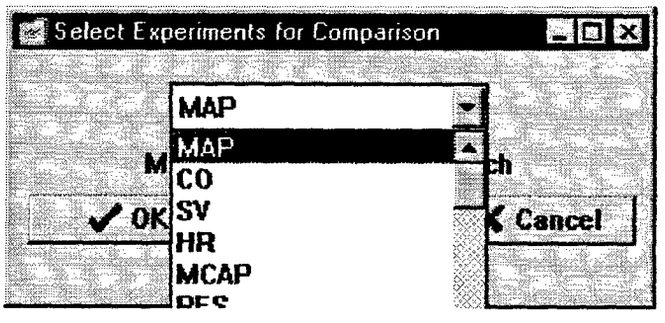
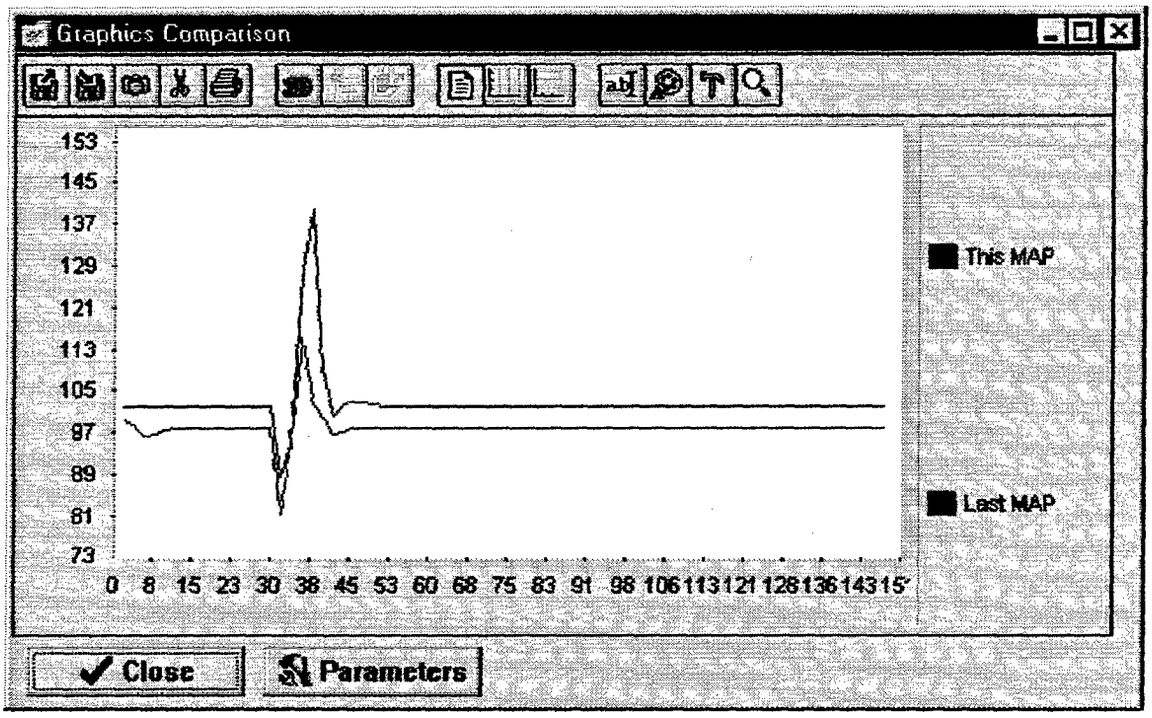
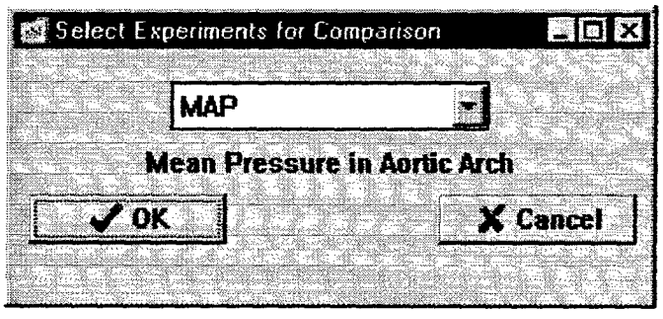


Fig. 25. Screen form for selection of variables to compare .



A.



B.

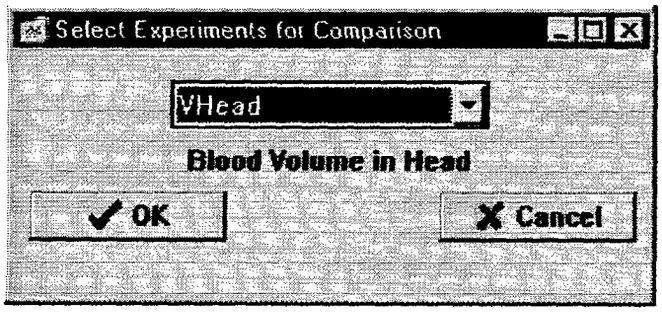


Fig.26. Graphs comparison (A).
The screen form illustrates that MAP (mean arterial pressure in aortic arch) was chosen for comparison (B).The next screen form (B) illustrates another situation when Vhead was chosen .

The "PILACCEL" allows also examine to all graphics in three-dimensional regime. This variant is illustrated for the case of comparison of blood volumes in head (VHead) for the current and the previous experiments.

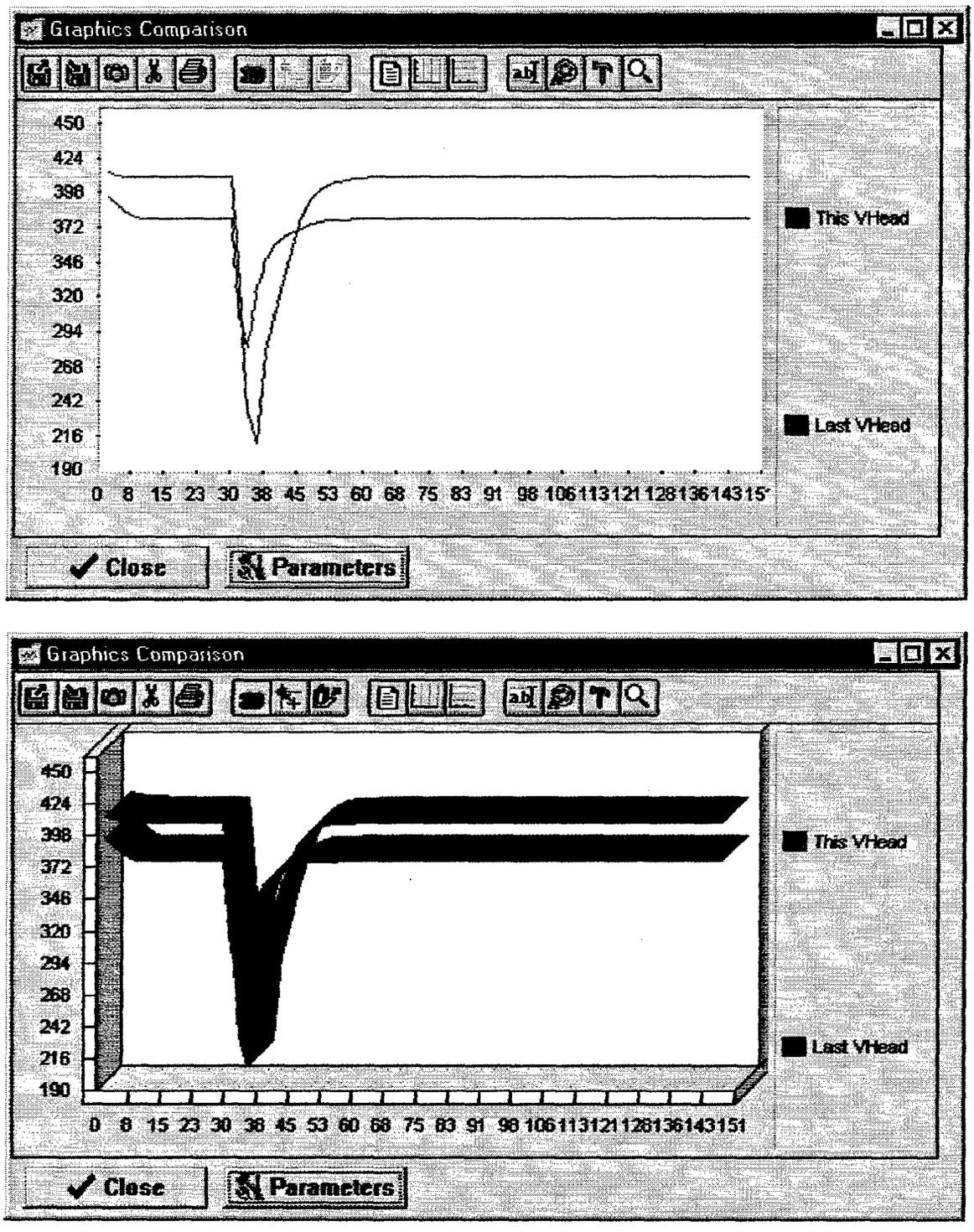


Fig.27. Graphs comparison in two- and three-dimensional regimes.

You can also see apply a coordinate net.

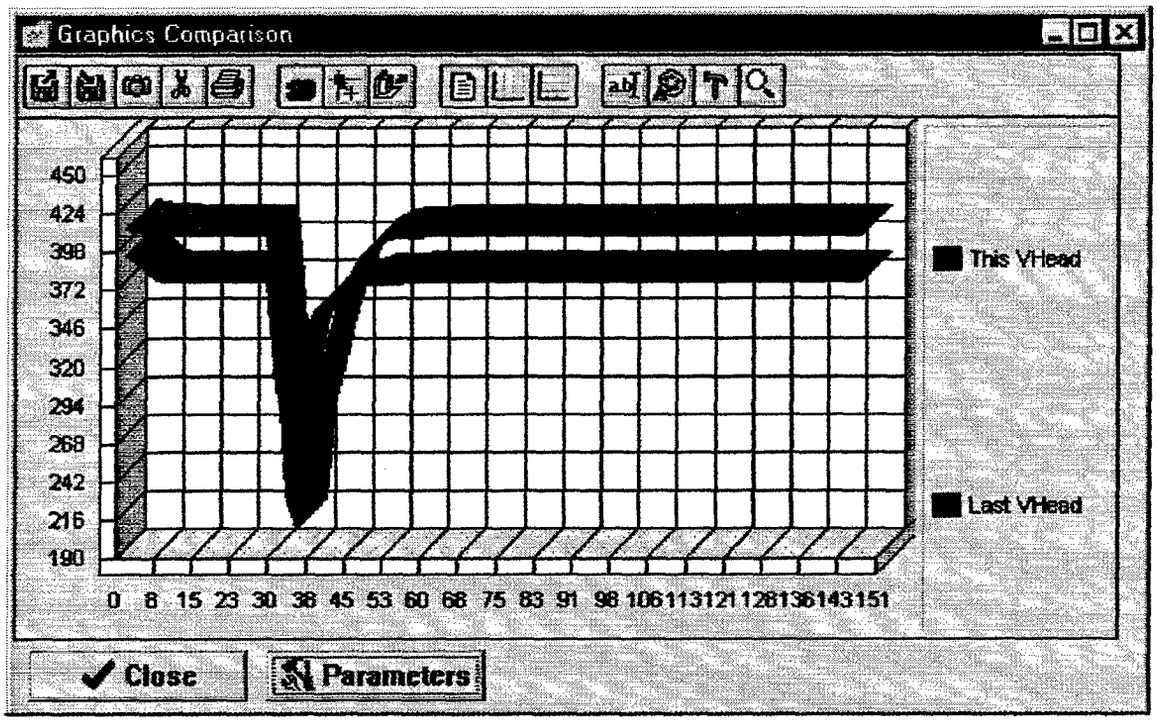


Fig.28. Comparison of graphs in three-dimensional reflection using coordinate net.

To get a hard copy of every experiment results as an experiment protocol (see fig.29) you should first use the last right speed button. The "PILACCEL" gives you an opportunity to write down your comments in this protocol (if you want to do it) and save them together. After that you can print the protocol.

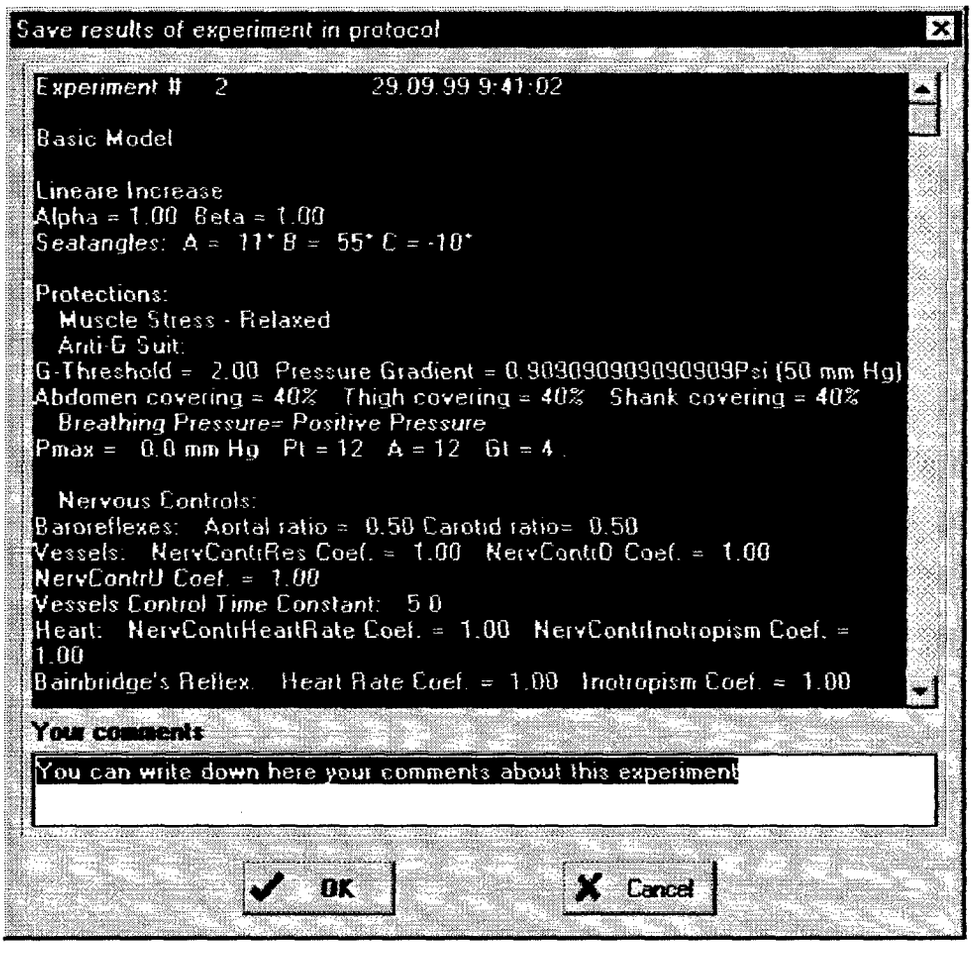


Fig.29. The screen form for saving the experiment results as a special Experiment protocol

The meaning of three speed-buttons located on the left side of the "PILACCEL" pannel, will be clear when you consult corresponding short help-windows. The left button serves as "Exit". The other two buttons serve for Experiment Options' saving (if you want to form some demo-options) and Experiment Options' loading. Using these buttons may be sometimes the shortest way to set slightly changed experiment parameters.

Chapter 5. DESCRIPTION OF MAIN TEST RESULTS

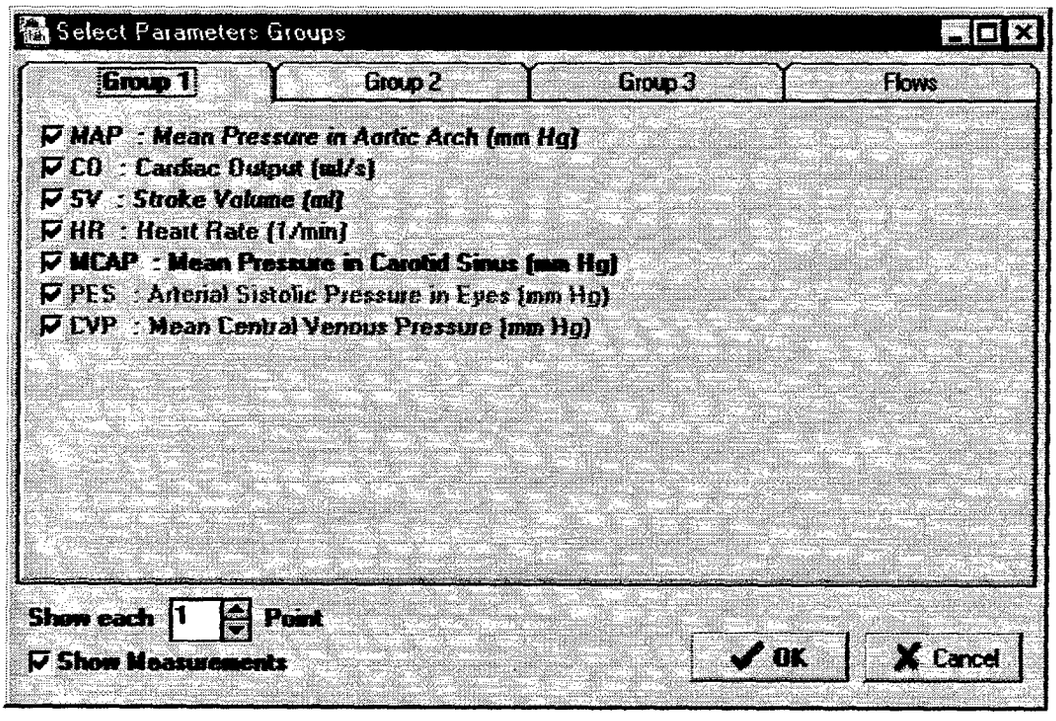
Models have been tested for a rather wide spectrum of situations. We are not able to present too much data in this report. We believe that the presentation of several basic regimes might help users to estimate the adequacy of models.

There are two different traditional approaches to analyze hemodynamics control mechanisms for gradual (usually 0.1g/s) and rapid (usually 1g/s or more) accelerations [1-5,14]. Keeping to this tradition we have presented in this report the test results for two acceleration regimes of simulation of hemodynamics for seated relaxed man. In the first case there are no artificial protective tools or methods, while in the second case there are combinations of different tools and methods.

As it has already been said in the previous chapter, every simulation experiment can be done after the user sets all information for this experiment, including experiment's maximal longitude if G-tolerance is present, i.e. until systolic arterial pressure in eye level becomes lower than 30 mm Hg. Just after this pressure get less of its control value, the software automatically generates transition from acceleration to the decceleration with the gradient of 0.25g/s.

According to our computational algorithms all calculations start from a given casual values of CVS's parameters. Therefore, some times there is transitional process that contains no useful information for user. To reach a stable hemodynamic regime for choosen first body position (horizontal or seated with different seat angles of legs and upper body part) we need some computational time. In fact, only after this time interval we have the "zero" time point to begin the real acceleration experiment. The end-point time to stop of elementary simulation experiment is the fixed time that we need to recover the starting phone stable hemodynamics after acceleration increasing was broken.

Figures 30A and 30B explain abbreviations for hemodynamics characteristics united into four different groups to facilitate their analysis.



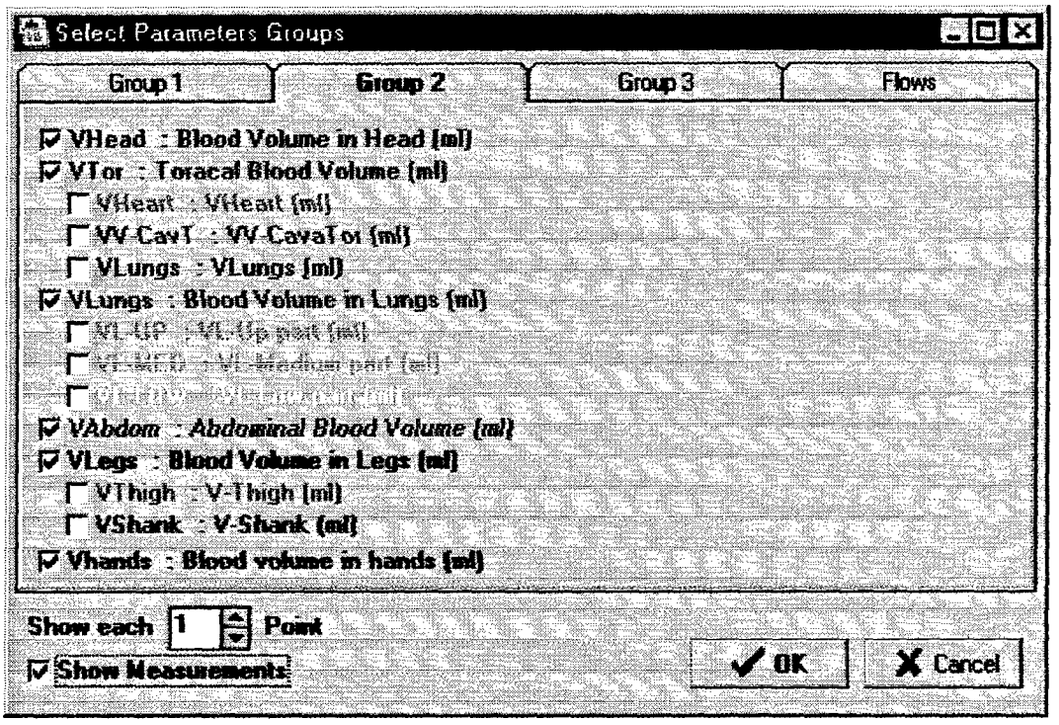
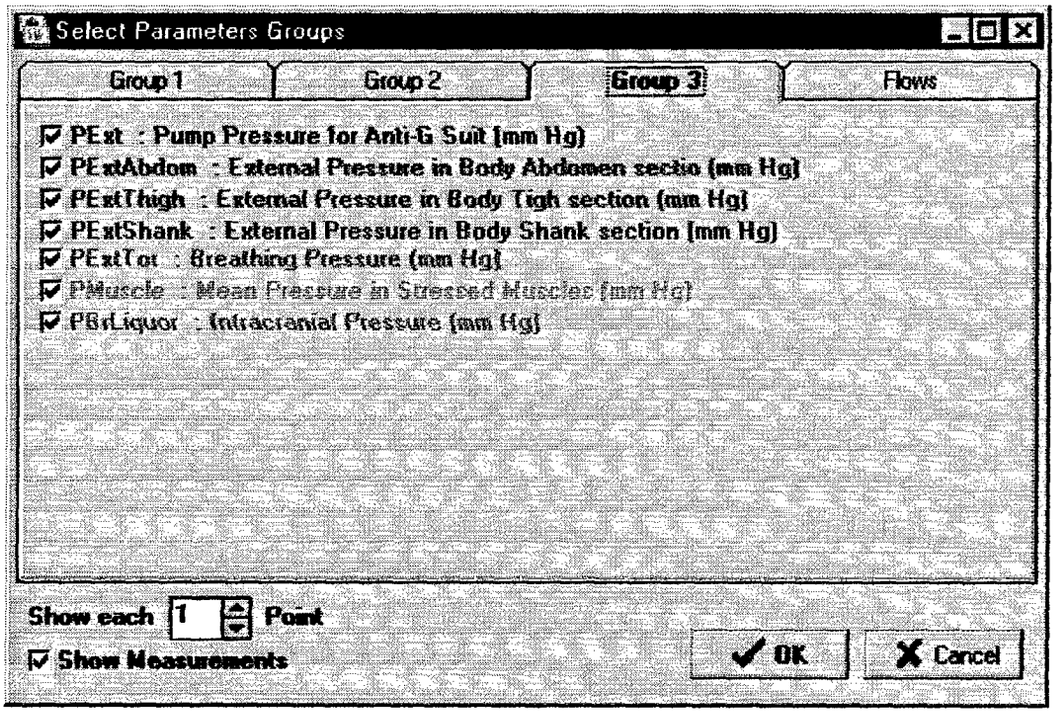


Figure 30A. Interpretation of abbreviations used on the figures 31-50.



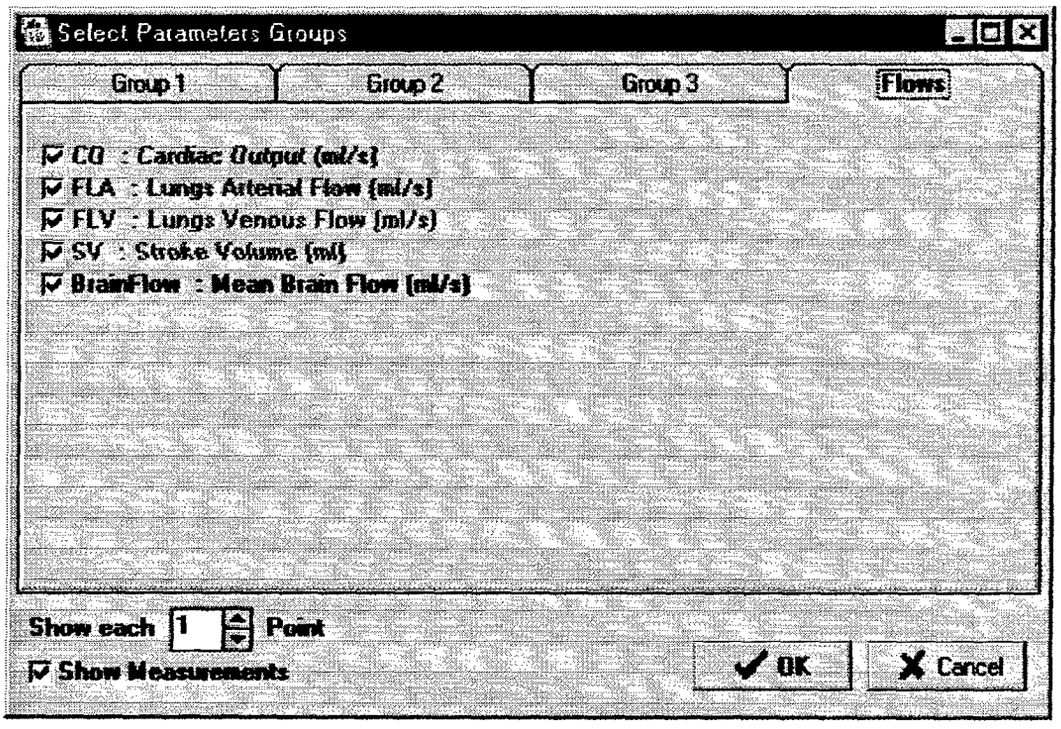


Figure 30B. Interpretation of abbreviations used on the figures 31-50.

Figures 31,32,33 and 34 illustrate dynamics of several groups of characteristics of central hemodynamics for simulation experiment which performed after the following simulation parameters have been set:

- standard seat-back angle 12 degrees ;
- G-onset is gradual 0.1g/s.

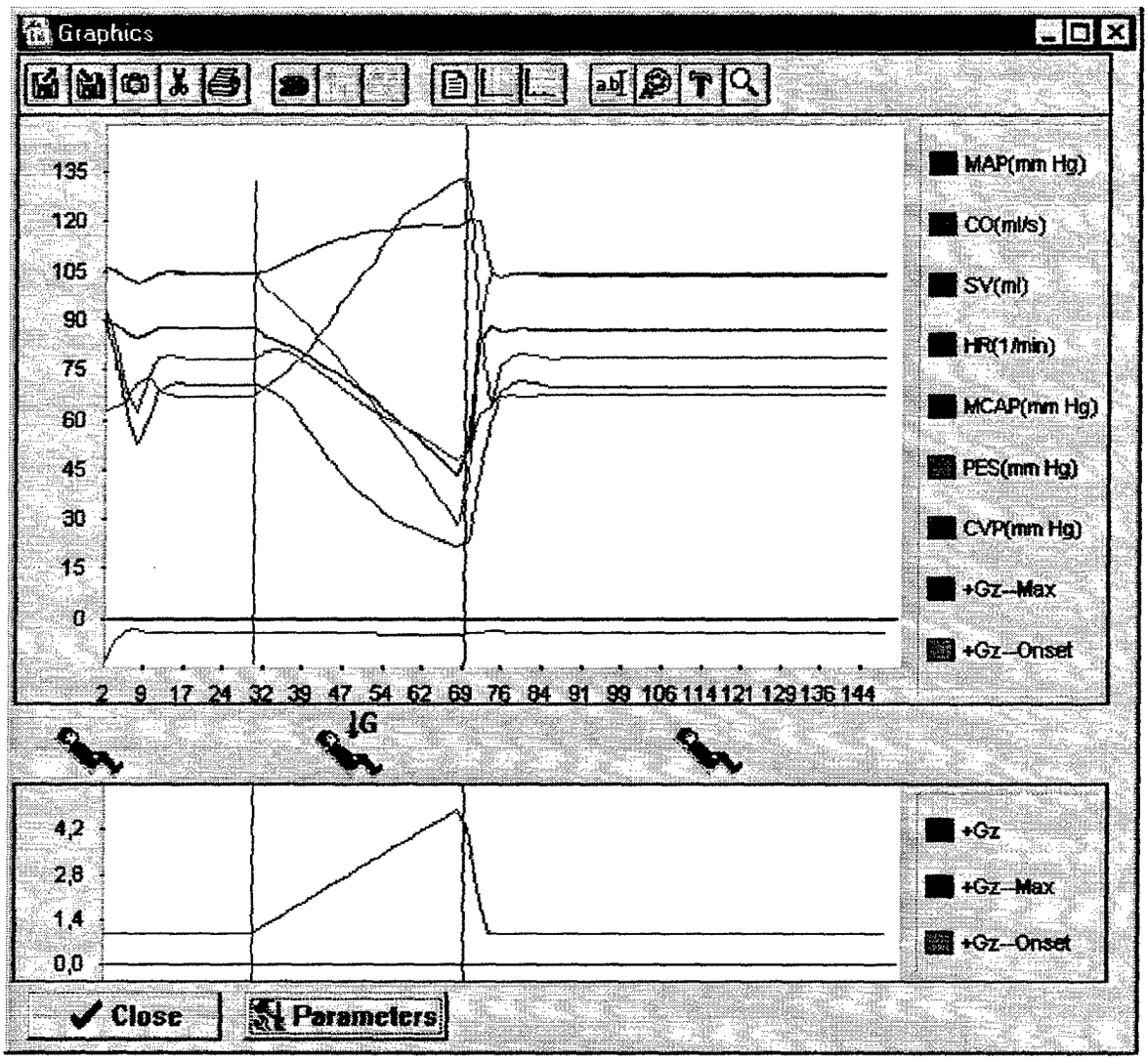


Figure 31. The dynamics of variables of CVS (above) and acceleration (below).

- standard seat-back angle 12 degrees ; G-onset is gradual 0.1g/s.
  (For the descriptions of abbreviations see fig.30).

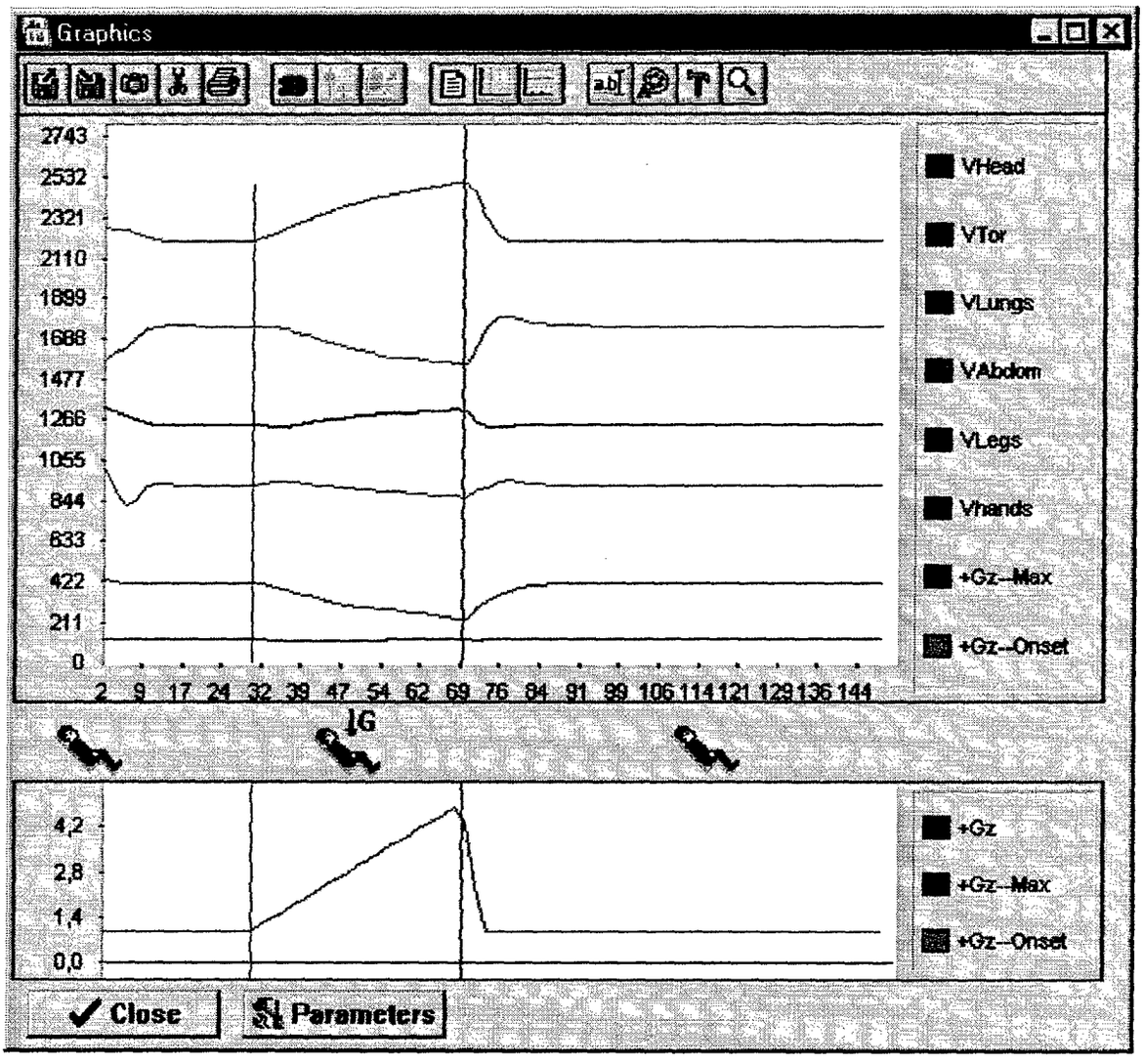


Figure 32. The dynamics of variables of CVS (above) and acceleration (below).

- standard seat-back angle 12 degrees ; G-onset is gradual 0.1g/s.
  (For the descriptions of abbreviations see fig.30).

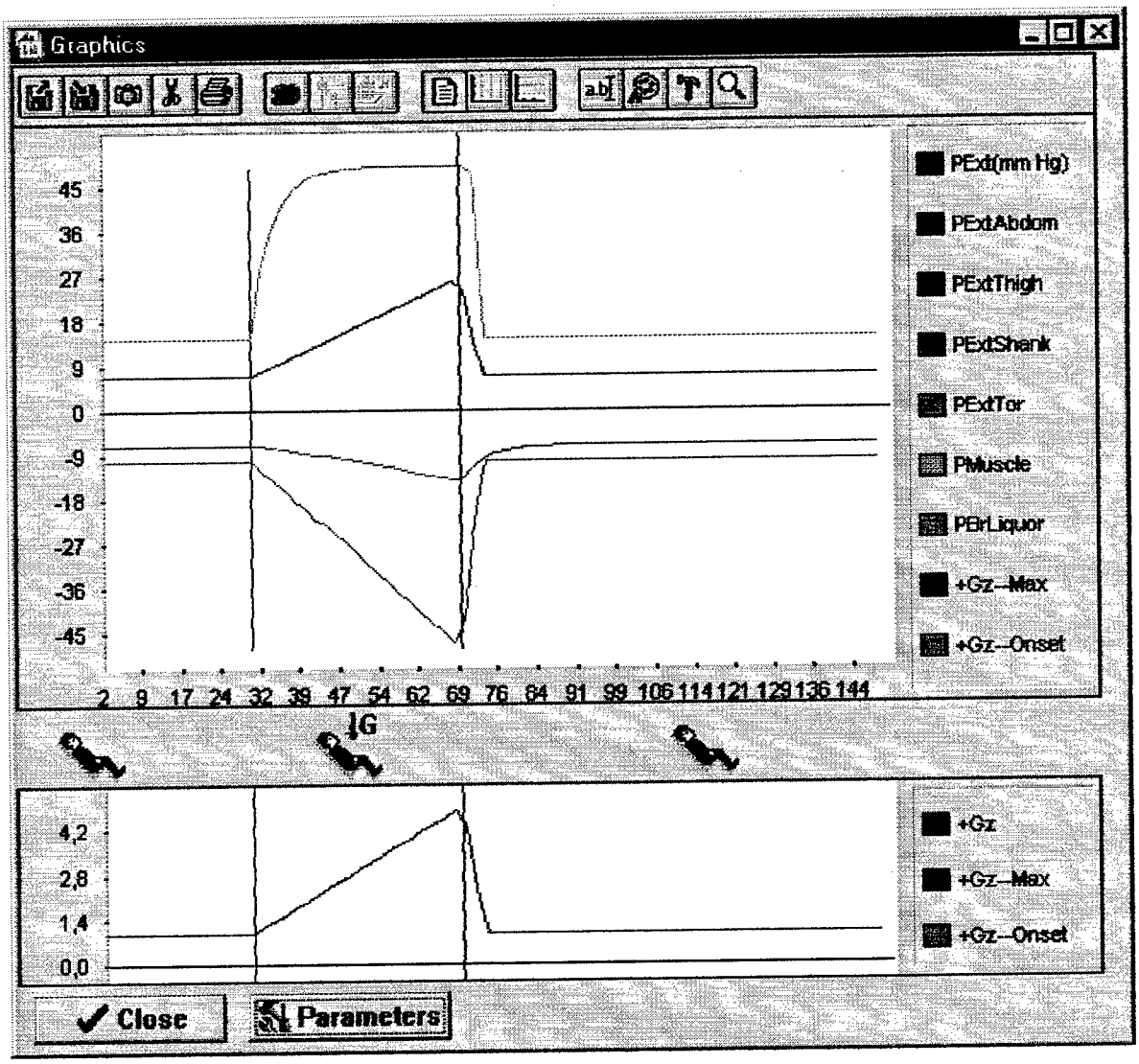


Figure 33. The dynamics of variables of CVS (above) and acceleration (below).
- standard seat-back angle 12 degrees ; G-onset is gradual 0.1g/s.
(For the descriptions of abbreviations see fig.30).

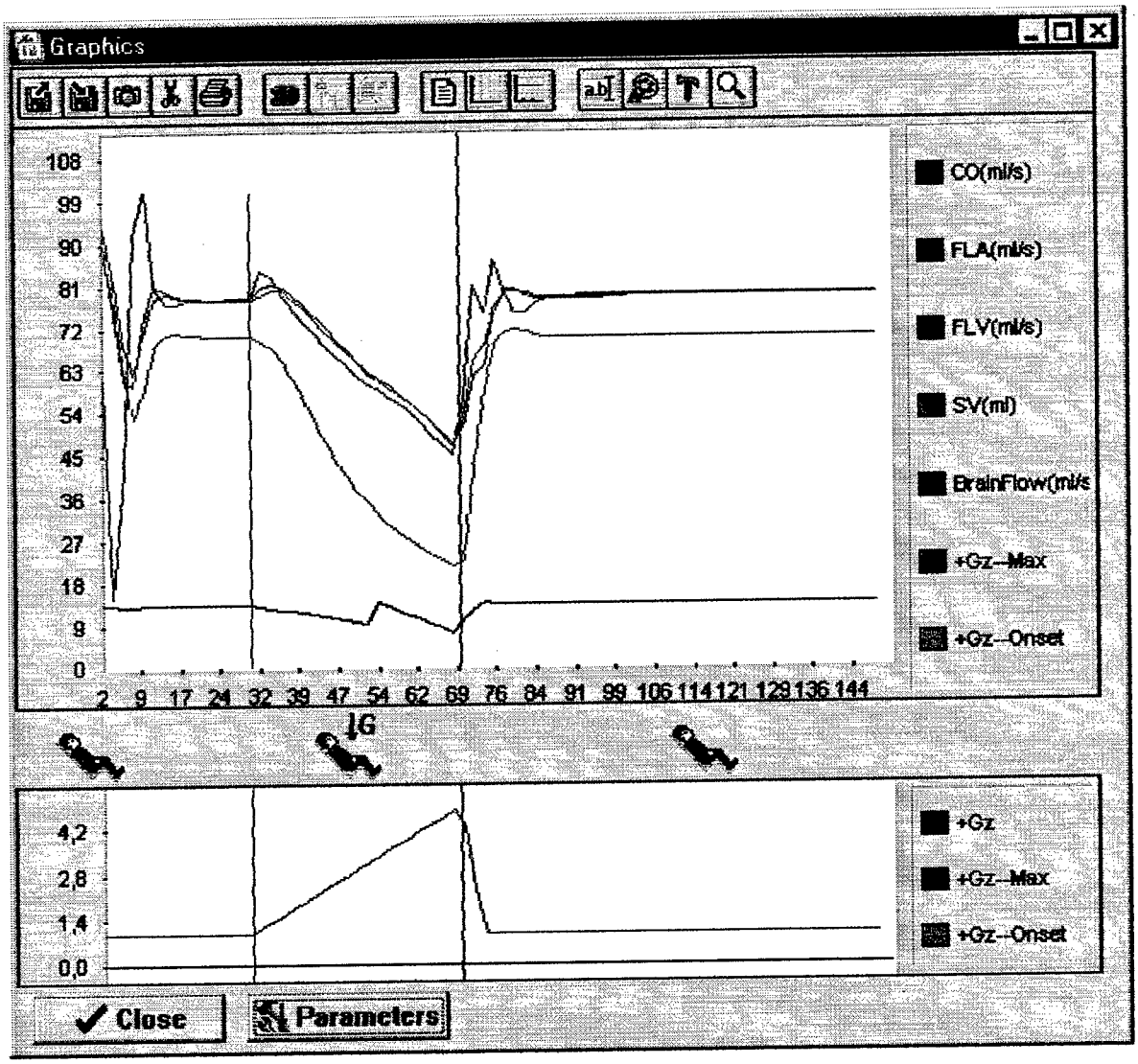


Figure 34. The dynamics of variables of CVS (above) and acceleration (below).

- standard seat-back angle 12 degrees ; G-onset is gradual 0.1g/s.
  (For the descriptions of abbreviations see fig.30).

Figures 35 and 36 illustrate dynamics of the same groups of central hemodynamics characteristics for simulation experiment carried out after the following simulation parameters have been set:

- standard seat-back angle 30 degrees ;
- G-onset is gradual 0.1g/s;

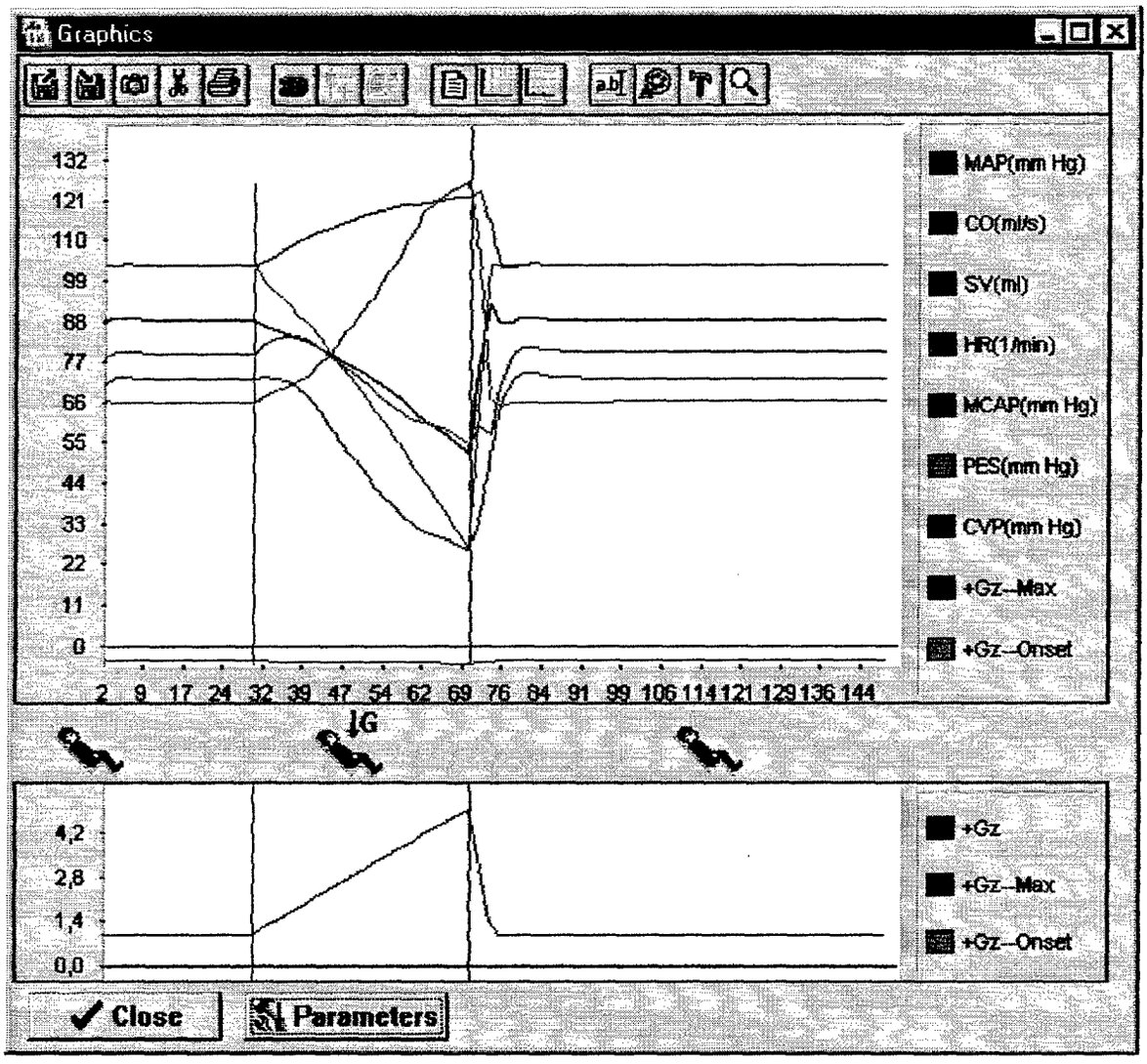


Figure 35. The dynamics of variables of CVS (above) and acceleration (below).

- standard seat-back angle 30 degrees ; G-onset is gradual 0.1g/s;
  (For the description of abbreviations see fig.30).

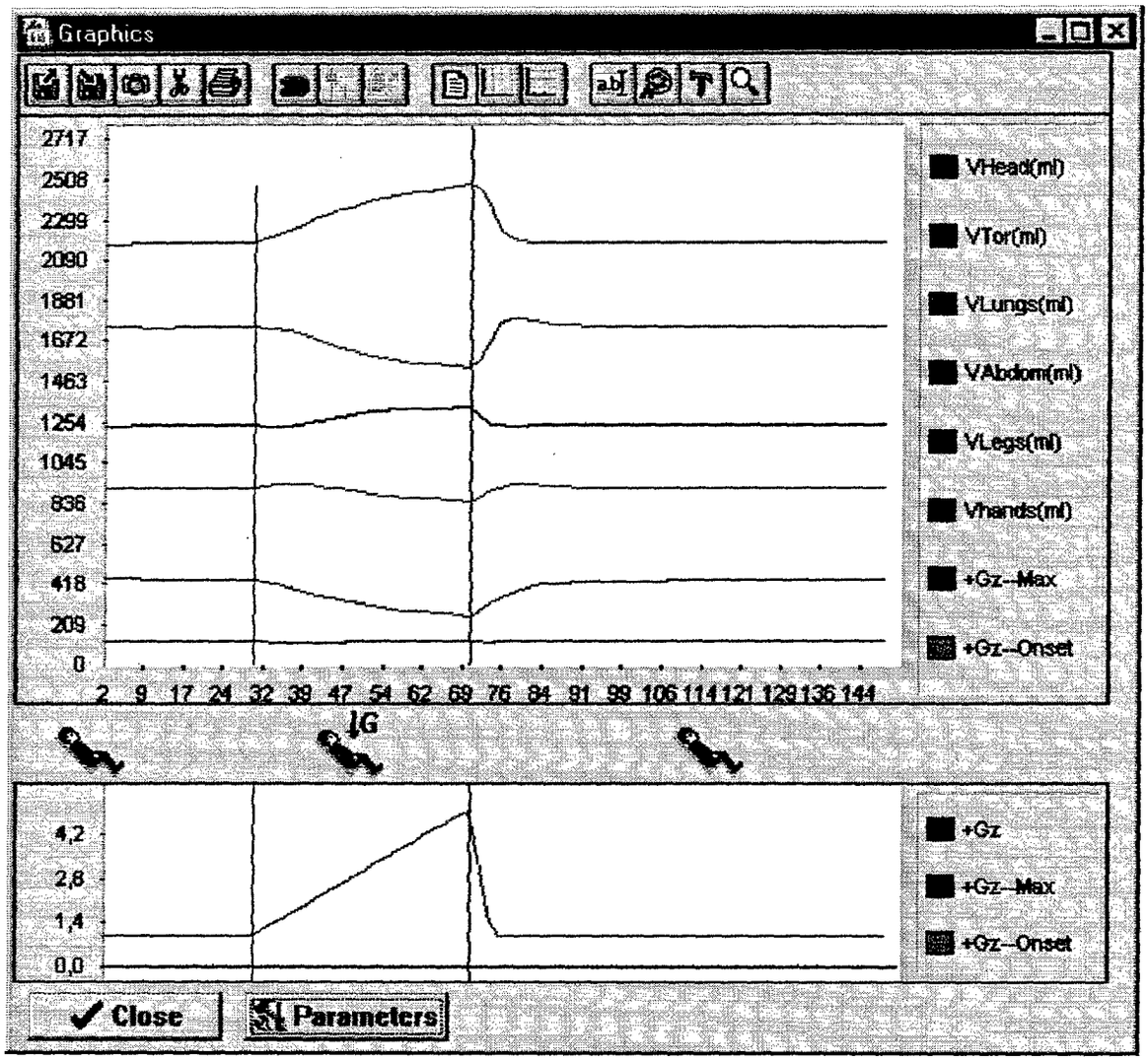


Figure 36. The dynamics of variables of CVS (above) and acceleration (below).

- standard seat-back angle 30 degrees ; G-onset is gradual 0.1g/s;
  (For the description of abbreviations see fig.30).

Figures 37 and 38 illustrate dynamics of the same groups of central hemodynamics characteristics for simulation experiment done after the following simulation parameters have been set:

- seat-back angle 65 grades ;
- G-onset is gradual 0.1g/s.

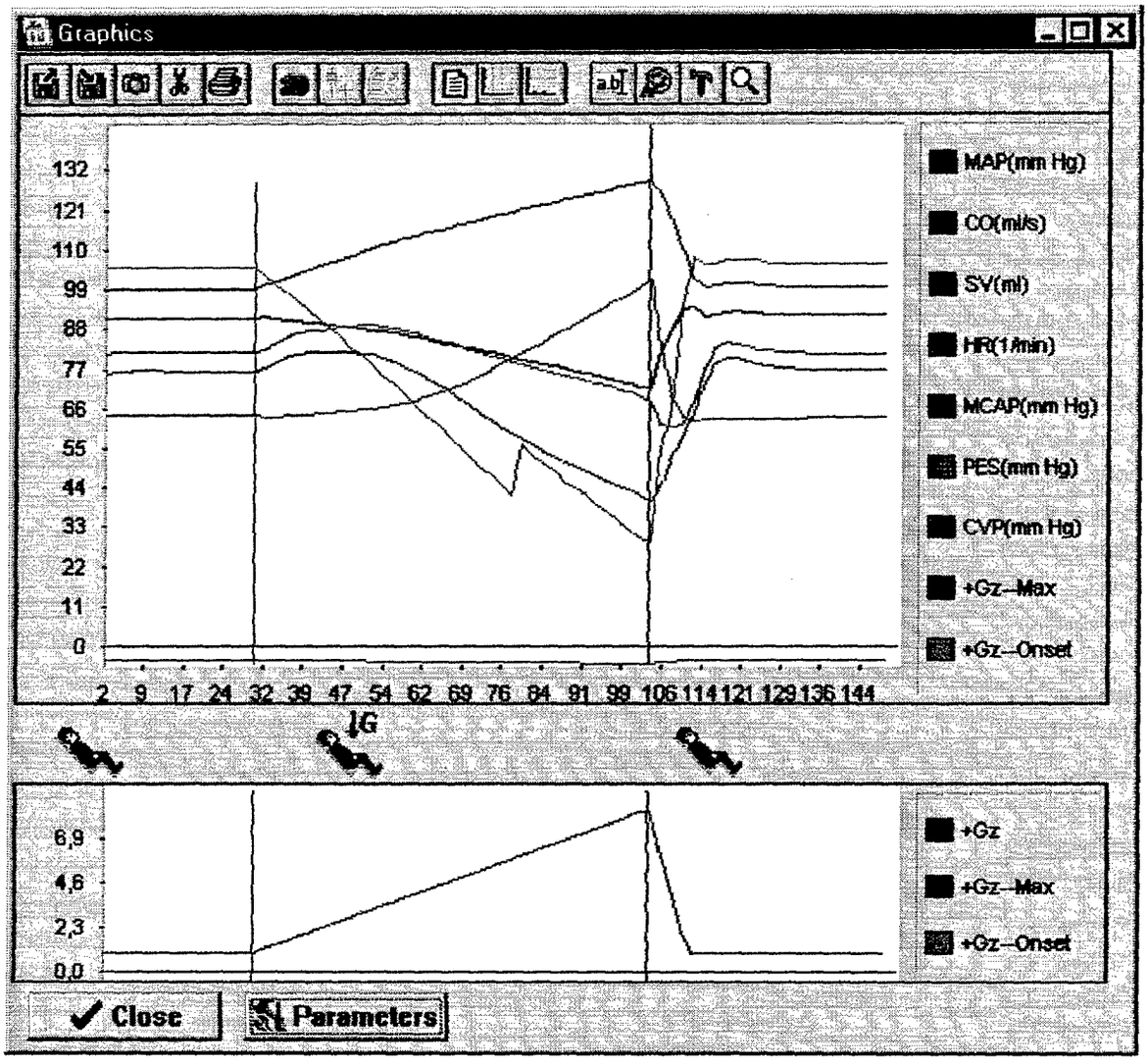


Figure 37. The dynamics of variables of CVS (above) and acceleration (below).

- seat-back angle 65 degrees ; G-onset is gradual 0.1g/s.
  (For the description of abbreviations see fig.30).

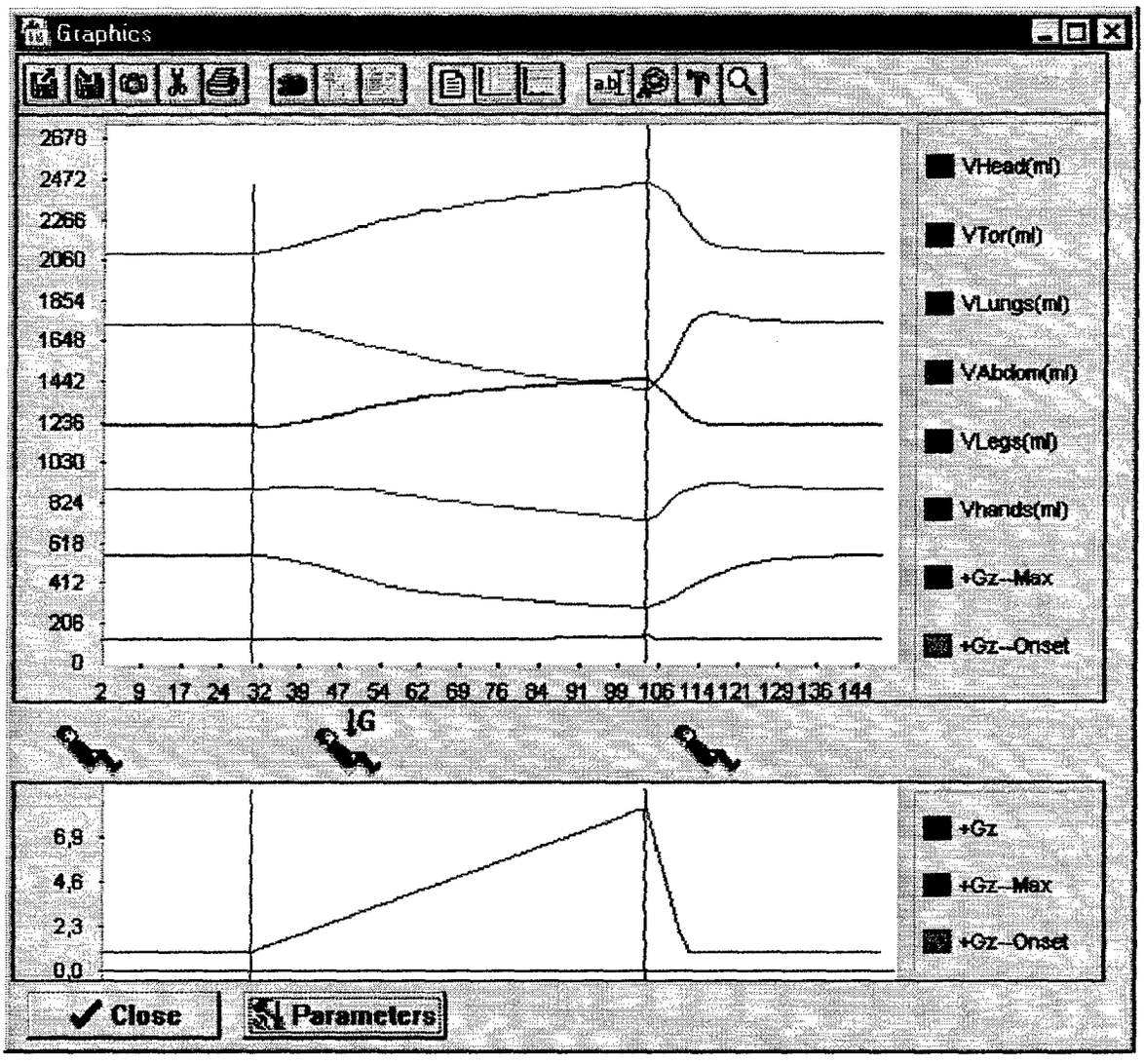


Figure 38. The dynamics of variables of CVS (above) and acceleration (below).

- seat-back angle 65 degrees ; G-onset is gradual 0.1g/s.
  (For the description of abbreviations see fig.30).

Figures 39, 40 and 41 illustrate the dynamics of the same groups of central hemodynamics characteristics for simulation experiment performed after the following simulation parameters have been set:

- seat-back angle 12 degrees ; G-onset is gradual 0.1g/s;
- Standard anti-G suit inflantation starts with 50 mm Hg / G after G-level reached 2.0G.

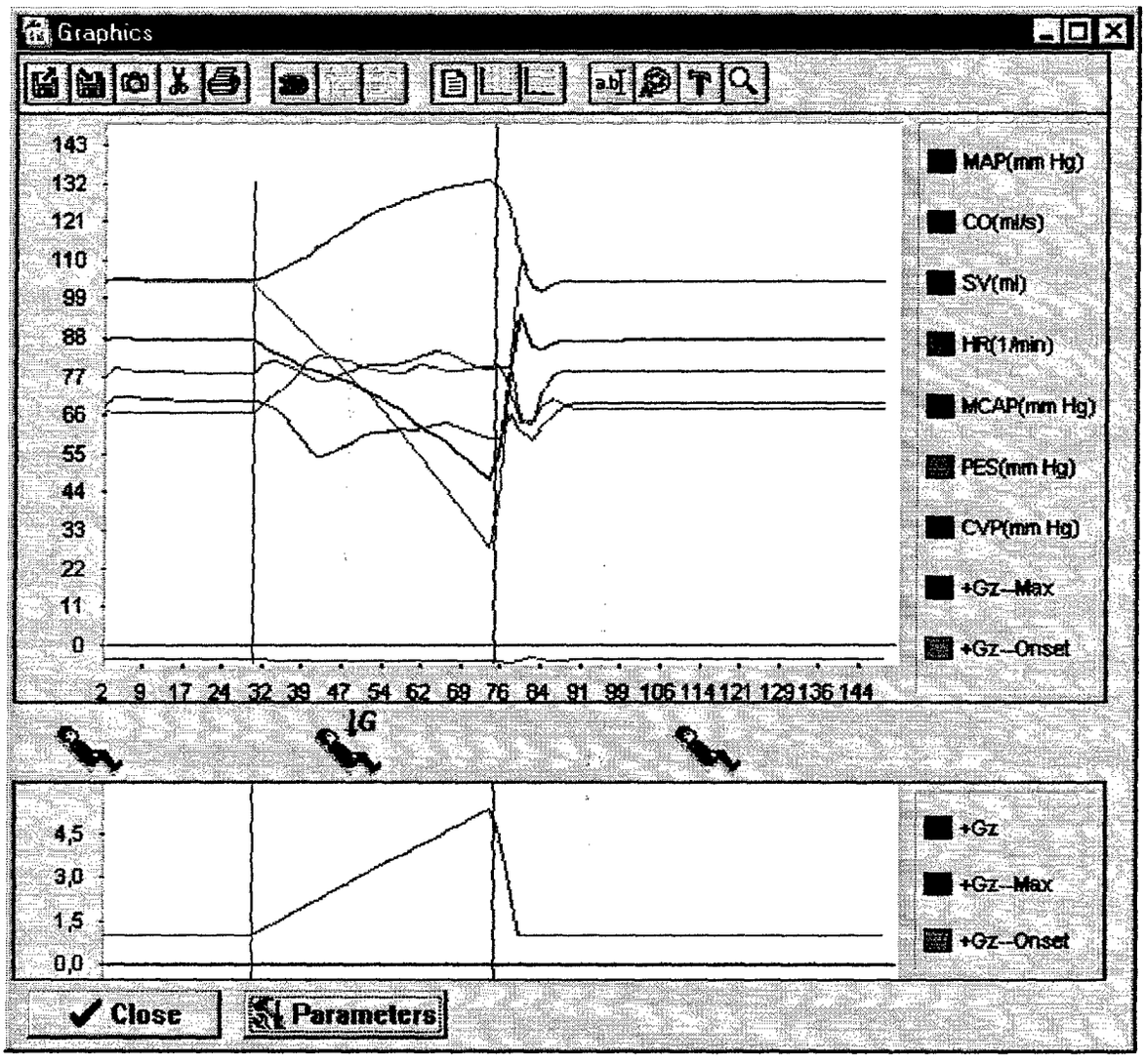


Figure 39. The dynamics of variables of CVS (above) and acceleration (below).

- seat-back angle 12 degrees ; G-onset is gradual 0.1g/s;
- Standard anti-G suit inflantation starts with 50 mm Hg / G after G-level reached 2.0G. (For the description of abbreviations see fig.30).

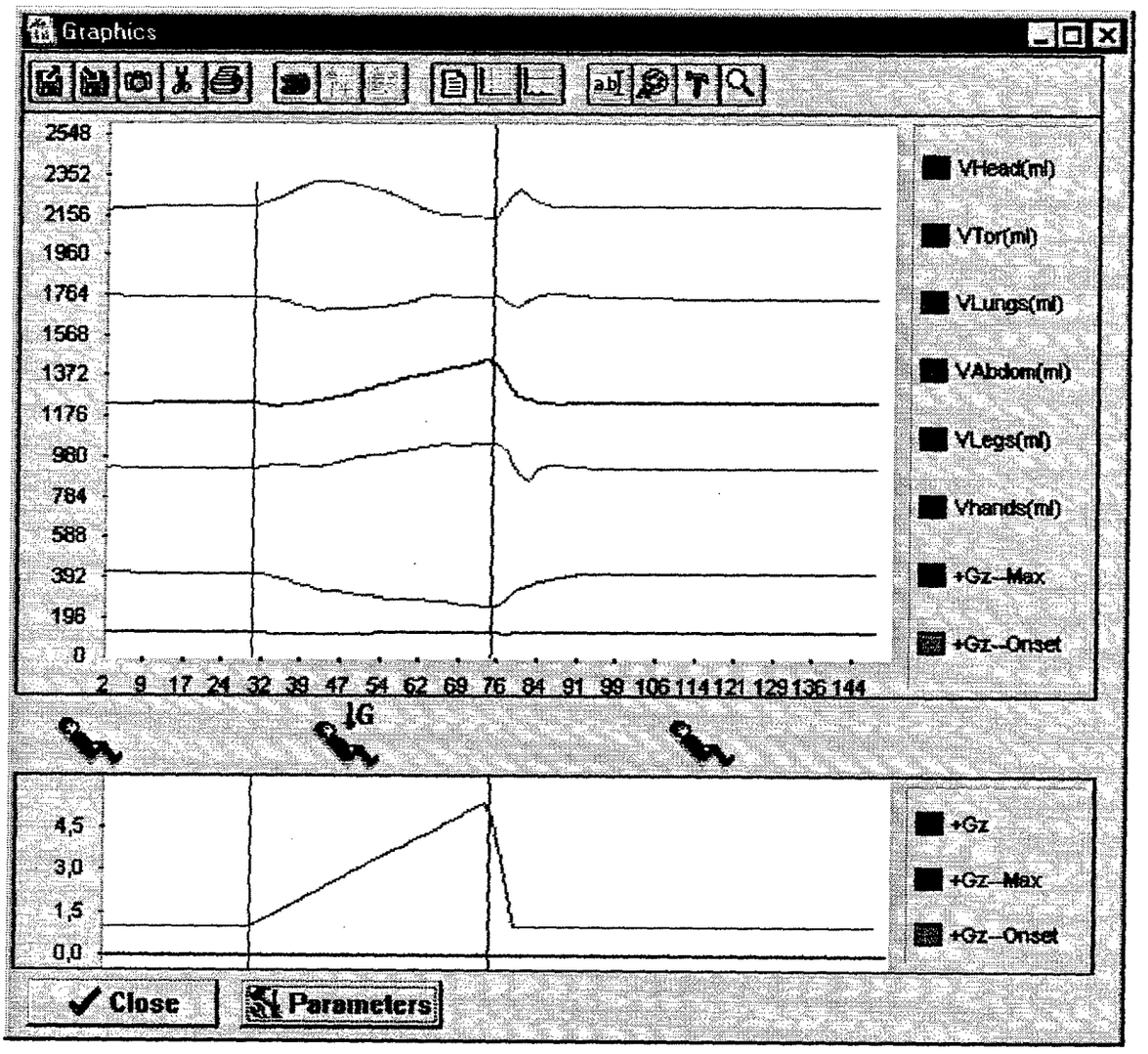


Figure 40. The dynamics of variables of CVS (above) and acceleration (low).

- standart seat-back angle 12 grades ; G-onset is gradual 0.1g/s;
- Standard anti-G suit inflantation starts with 50 mm Hg / G after G-level reached 2.0G. (For the description of abbreviations see fig.30).

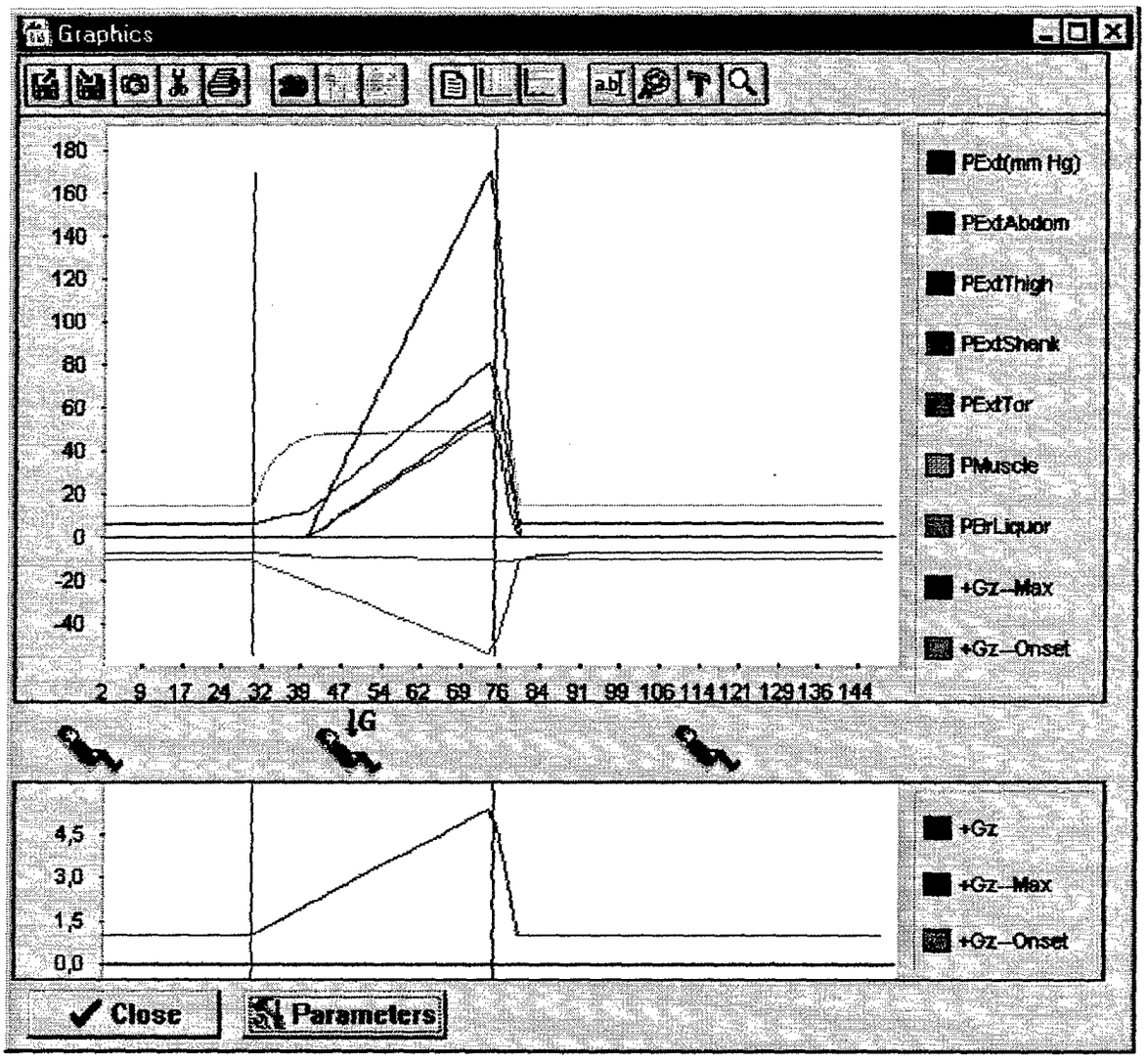


Figure 41. The dynamics of variables of CVS (above) and acceleration (below).

- seat-back angle 12 grades ; G-onset is gradual 0.1g/s;
- Standard anti-G suit inflantation starts with 50 mm Hg / G after G-level reached 2.0G.

(For the description of abbreviations see fig.30).

Figures 42,43 and 44 illustrate dynamics of the same groups of central hemodynamics characteristics for simulation experiment performed setting the following simulation parameters:
seat-back angle 12 grades ;G-onset is gradual 0.1g/s;
Standard anti-G suit inflantation starts with 50 mm Hg / G after G-level reached 2.0G.

After started Positive pressure breathing (PPB) with gradient of 12 mm Hg / G starts after G-level reached 4.0G and until G-level is less than 60 mm Hg.

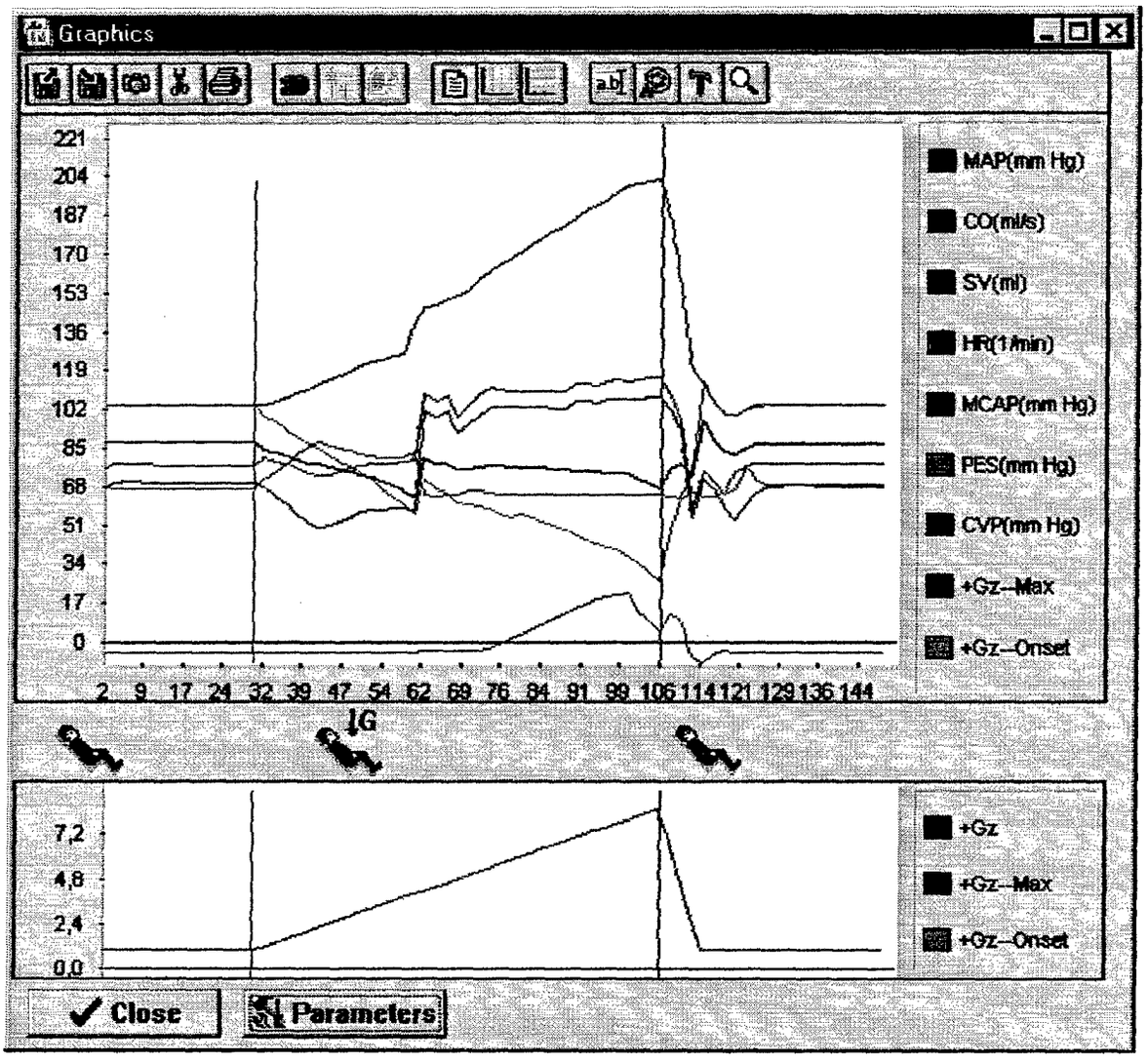


Figure 42. The dynamics of variables of CVS (above) and acceleration (below).

- seat-back angle 12 grades ;G-onset is gradual 0.1g/s;
- Standard anti-G suit inflantation starts with 50 mm Hg / G after G-level reached 2.0G.
- After started Positive pressure breathing (PPB) with gradient of 12 mm Hg / G starts after G-level reached 4.0G and until G-level is less than 60 mm Hg.

(For the description of abbreviations see fig.30).

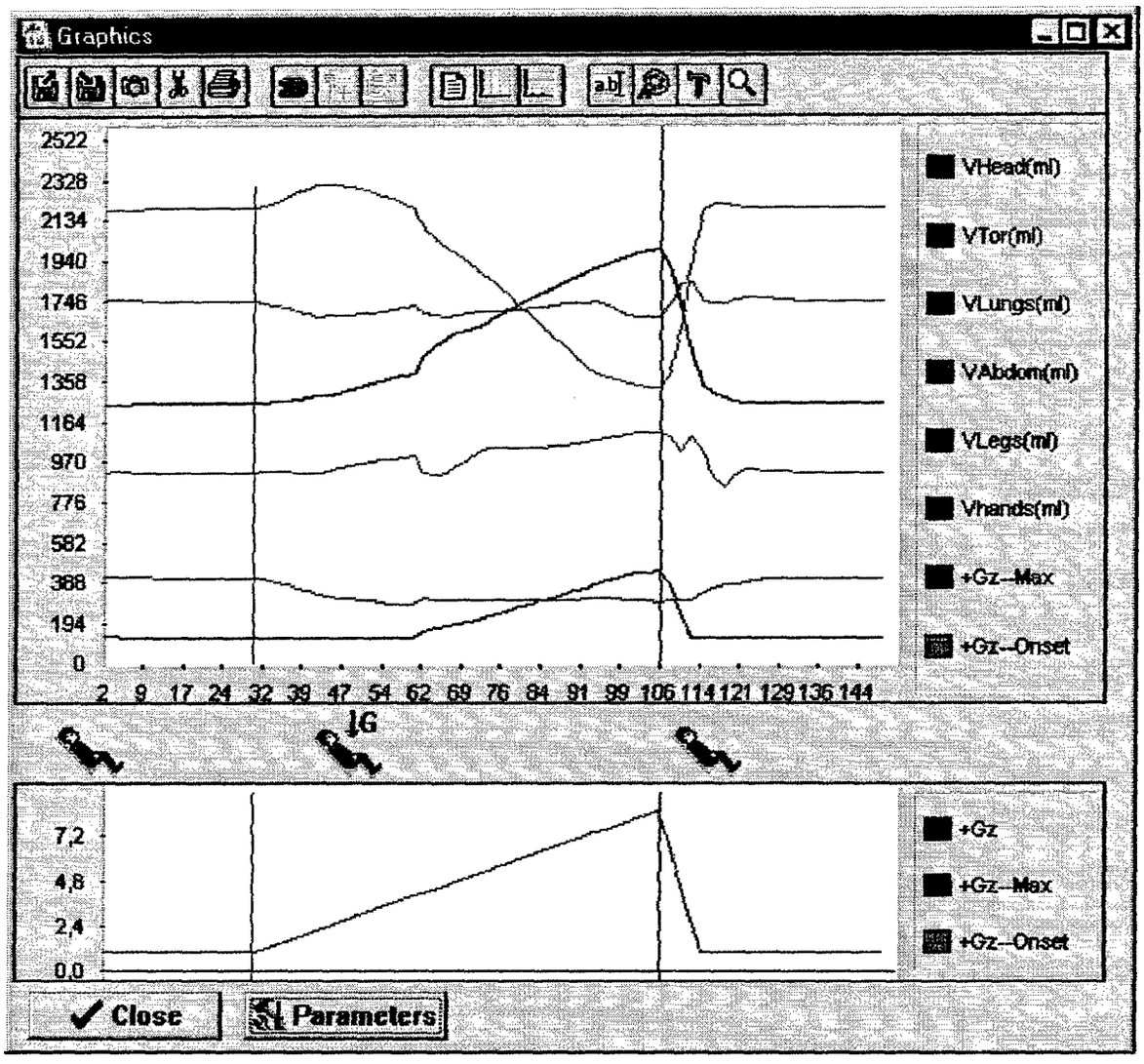


Figure 43. The dynamics of variables of CVS (above) and acceleration (below).

- seat-back angle 12 grades ;G-onset is gradual 0.1g/s;
- Standard anti-G suit inflantation starts with 50 mm Hg / G after G-level reached 2.0G.
- After started Positive pressure breathing (PPB) with gradient of 12 mm Hg / G starts after G-level reached 4.0G and until G-level is less than 60 mm Hg.

(For the description of abbreviations see fig.30).

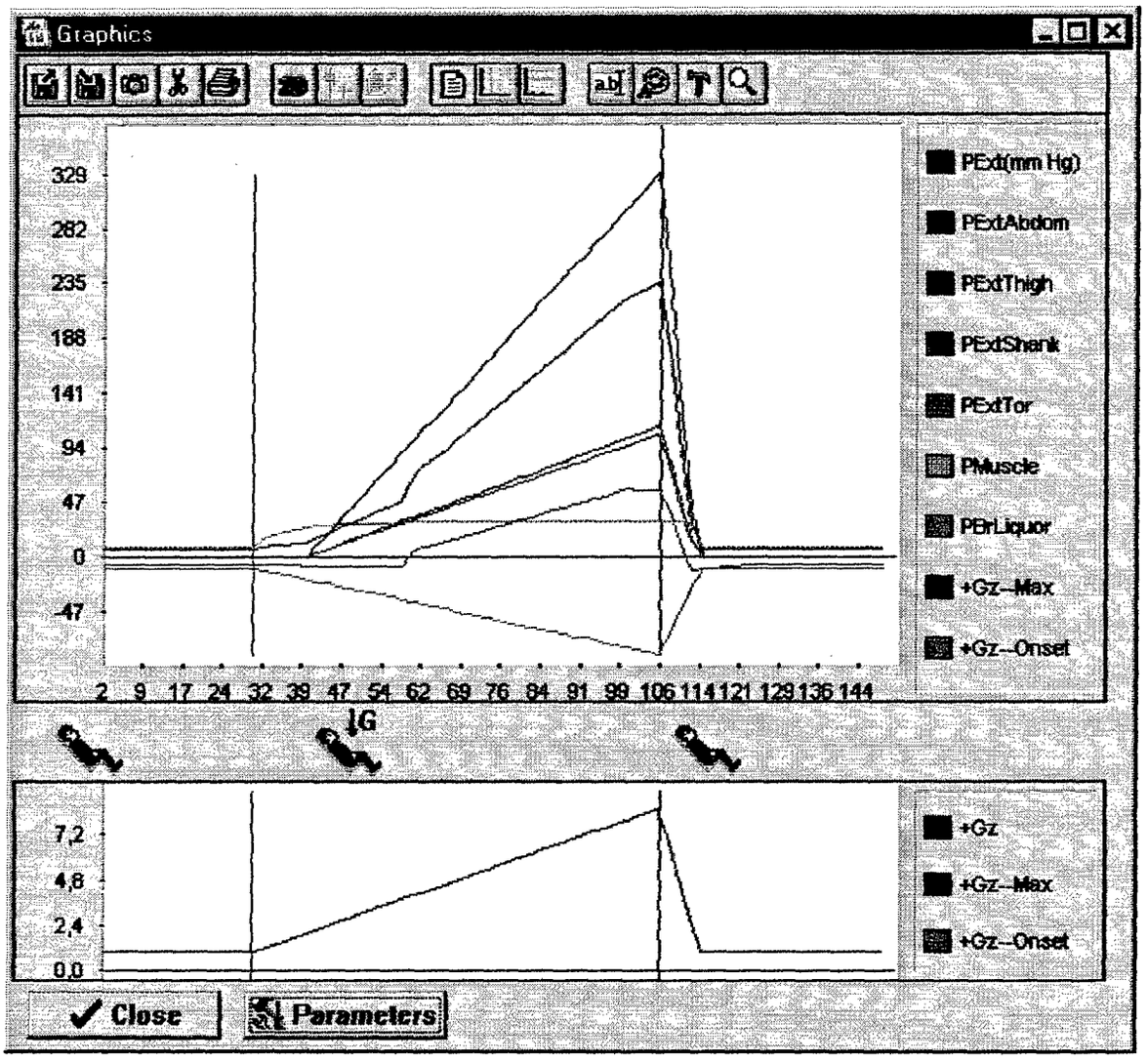


Figure 44. The dynamics of variables of CVS (above) and acceleration (below).

- seat-back angle 12 grades ;G-onset is gradual 0.1g/s;
- Standard anti-G suit inflantation starts with 50 mm Hg / G after G-level reached 2.0G.
- After started Positive pressure breathing (PPB) with gradient of 12 mm Hg / G starts after G-level reached 4.0G and until G-level is less than 60 mm Hg.
  (For the description of abbreviations see fig.30).

The next group of illustrations is related to the simulation regime for rapid accelerations with 1.0g/s. We presented only two situations: first (figures 45,46 and 47)- relaxed muscles without any artificial protection, and second (figures 48, 49 and 50)- with full protection, i.e. anti-G suit, muscle stress, PPG. The seat-back angle was equal to 12 degrees.

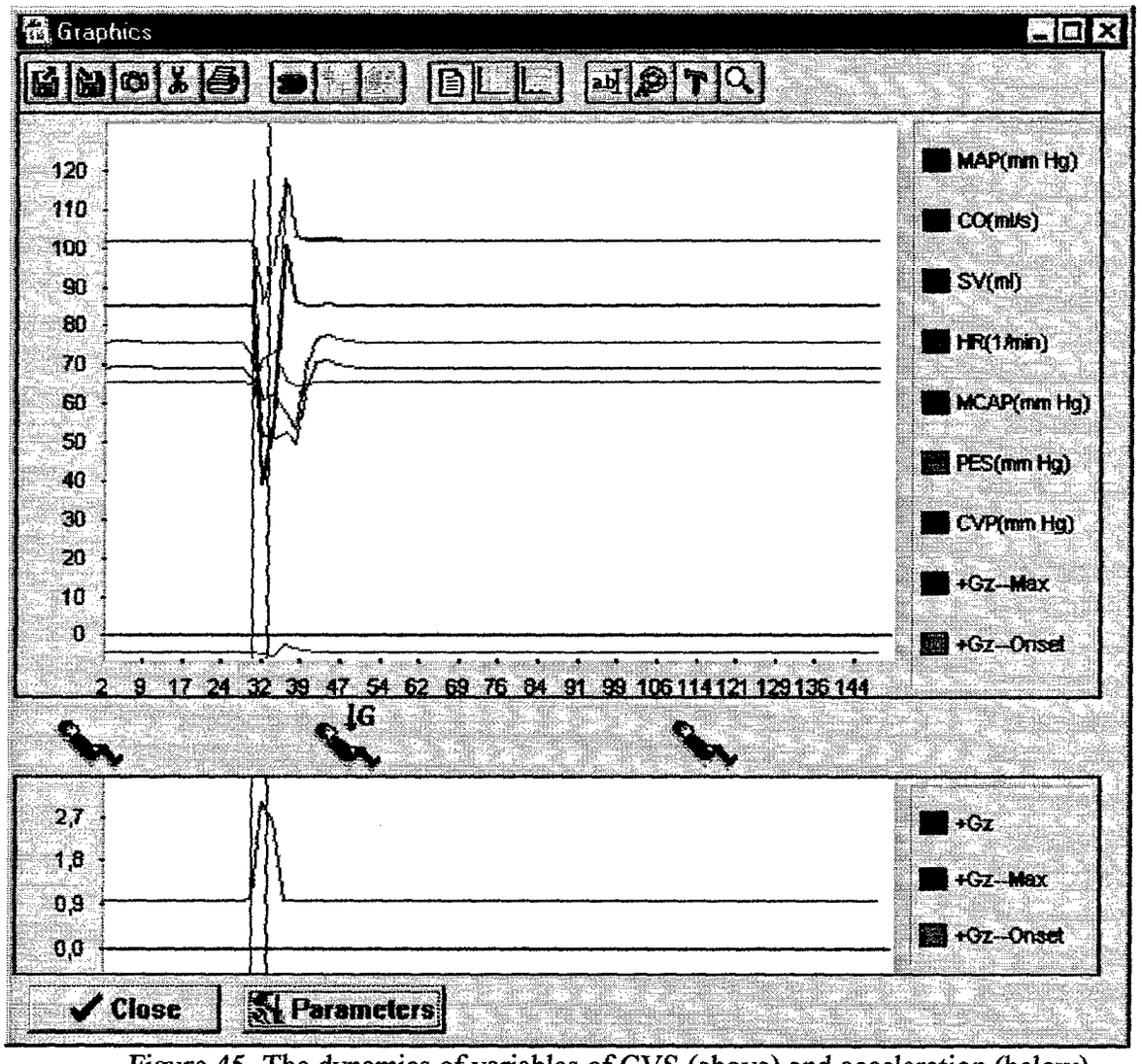


Figure 45. The dynamics of variables of CVS (above) and acceleration (below).

- Without protections: relaxed muscles.
  (For description of abbreviations see fig.30).

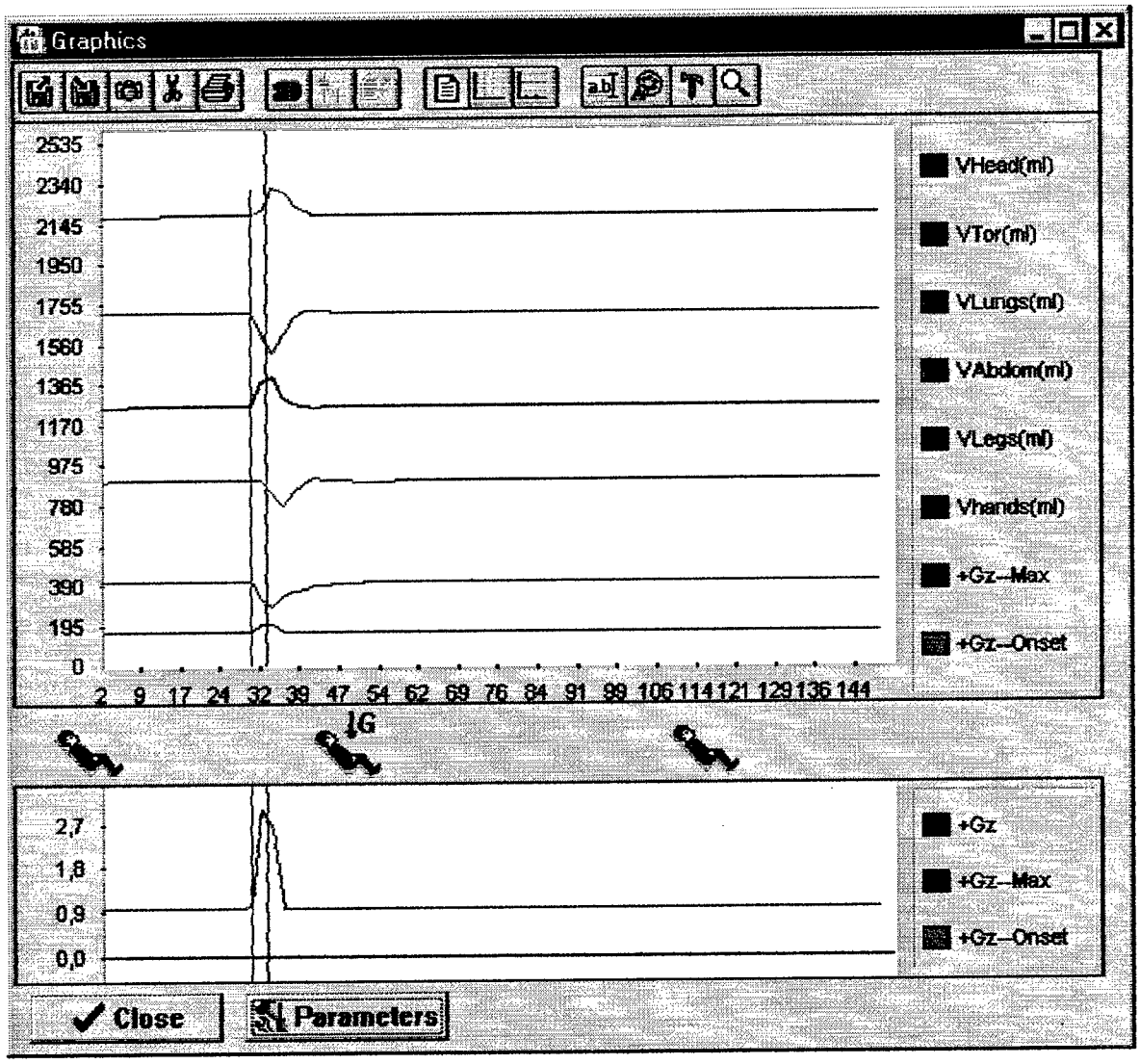


Figure 46. The dynamics of variables of CVS (above) and acceleration (below).

- Without protections: relaxed muscles.
  (For description of abbreviations see fig.30).

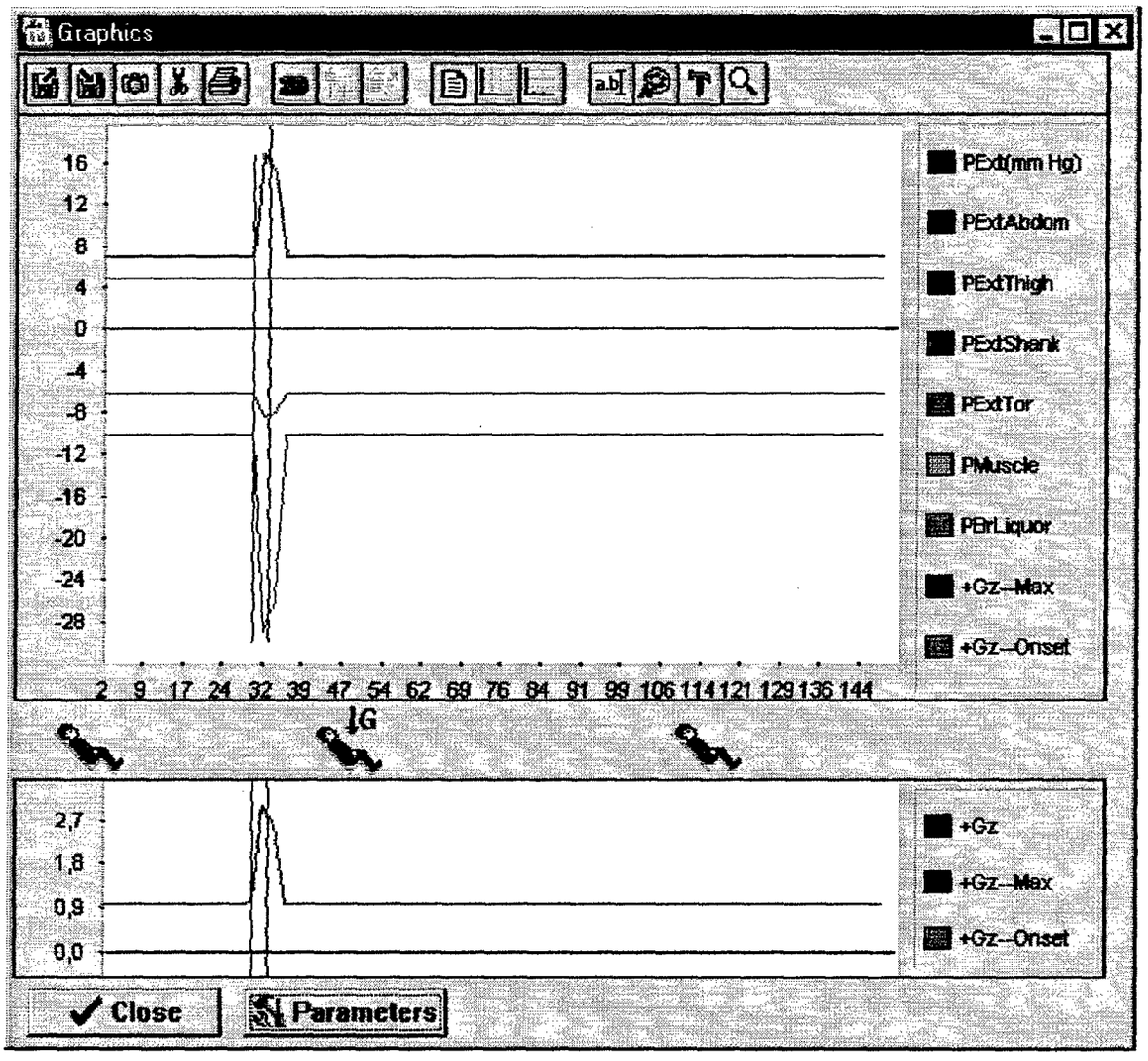


Figure 47. The dynamics of variables of CVS (above) and acceleration (below).

- Without protections: relaxed muscles.
  (For description of abbreviations see fig.30).

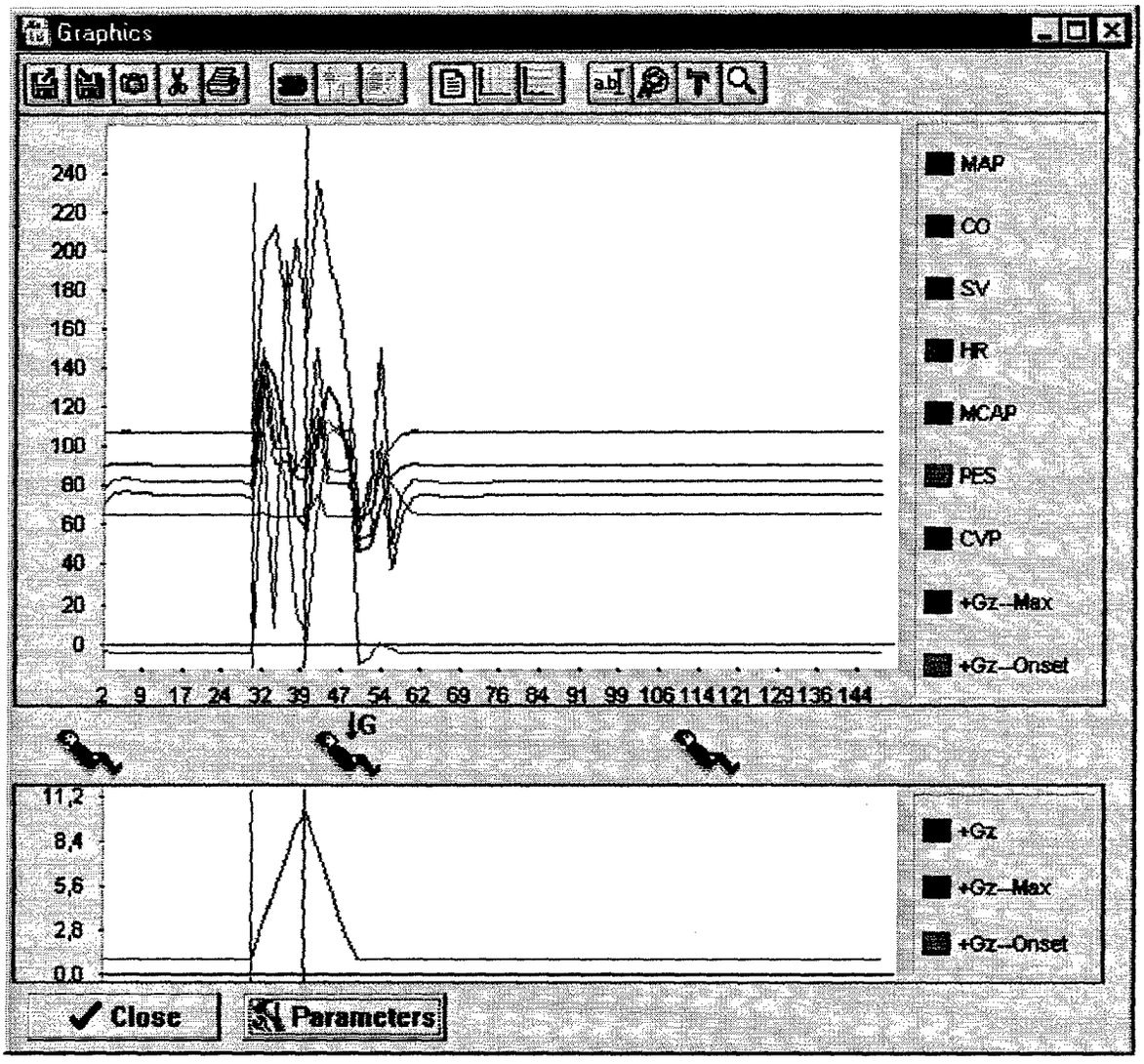


Figure 48. The dynamics of variables of CVS (above) and acceleration (below).

- The seat-back angle 12 degrees.
- Full protection.

(For the description of abbreviations see fig.30).

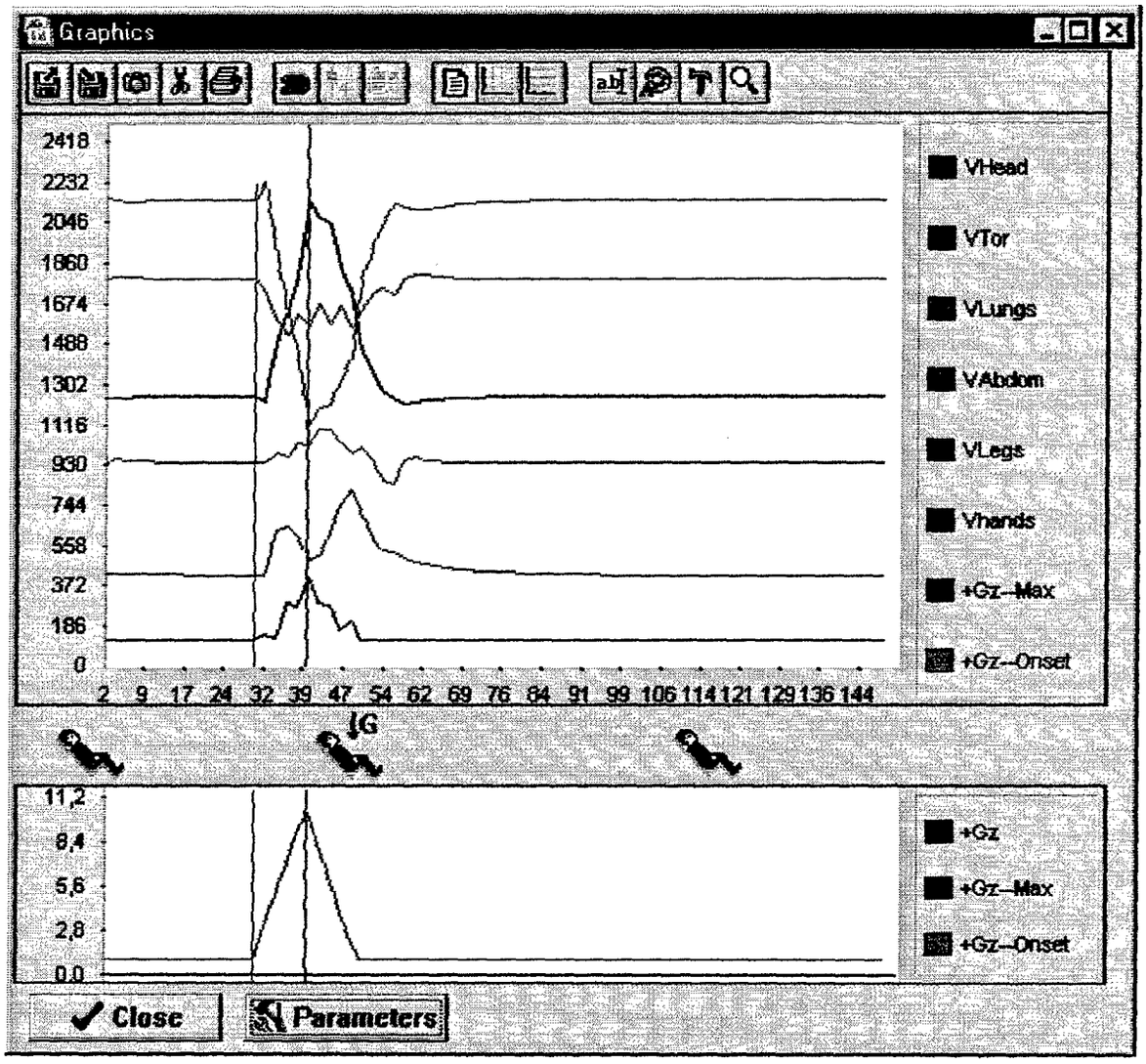


Figure 49. The dynamics of variables of CVS (above) and acceleration (below).

- The seat-back angle 12 degrees.
- Full protection.

(For the description of abbreviations see fig.30).

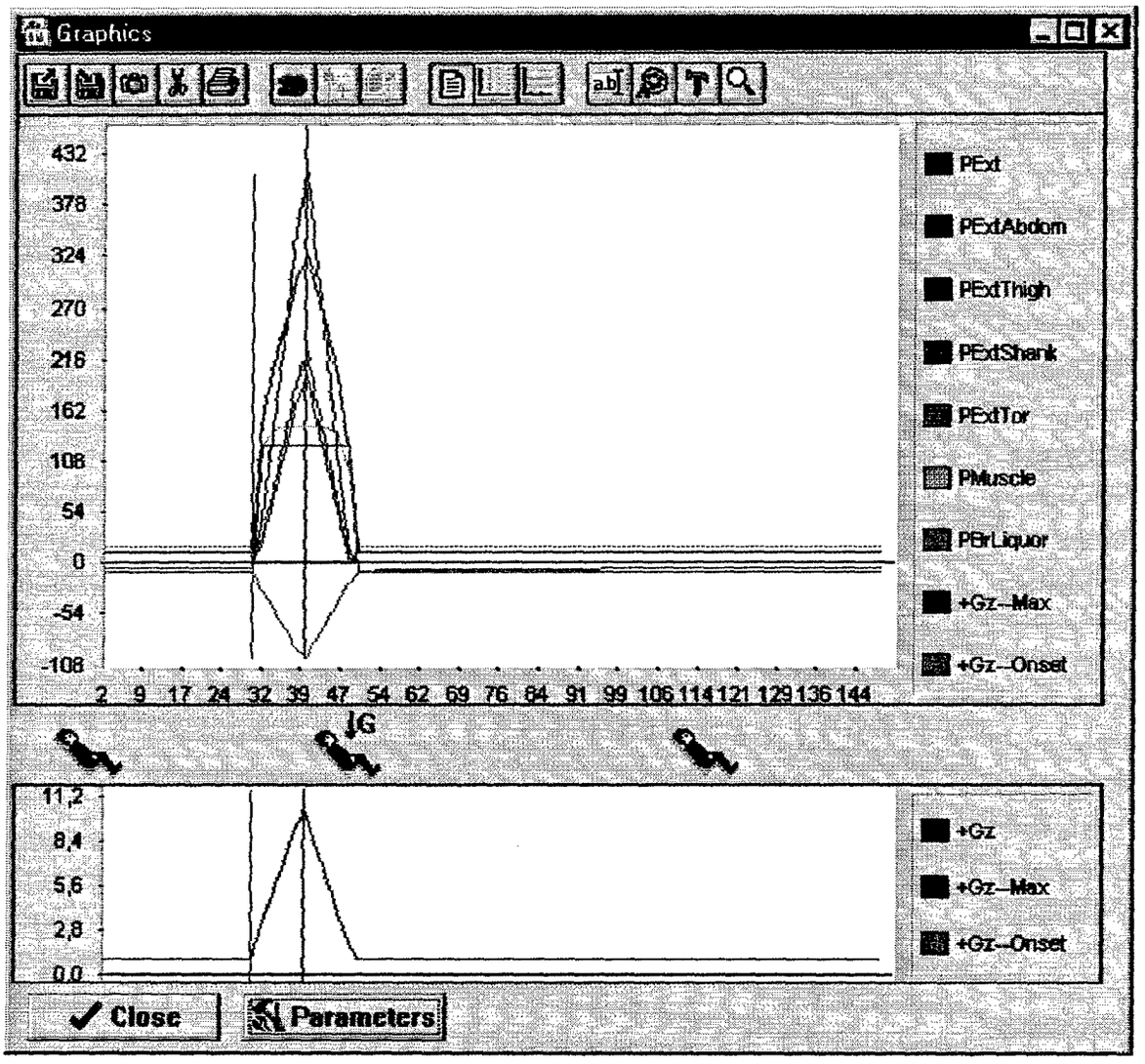


Figure 50. The dynamics of variables of CVS (above) and acceleration (below).

- The seat-back angle 12 degrees.
- Full protection.

(For the description of abbreviations see fig.30).

DISCUSSION

Presented examples of test simulation results may be considered only as minimal illustrative part of large computer experiments that were done to find the adequacy quantitative values for the main constants of models. We think, these illustrations are representative to show readers the adequacy of models for main known situations. Much more simulation situations were analyzed during testing of models in WPAFB. At the same time we would like to note that there are not real investigations presenting similar quantitative data to check all of model testing results. Experimentators usually have only a little part of information about human hemodynamics. Having our software they could be able compare their limited data with the similar simulation data. If there is an acceptable accordance between these data the experimentator should assume that all additional simulation data are also reliable. But we are not sure that this assertion is always confident.

The developed variant of informational technology does not permit users to access the inside of models to modify their different parameters. So, although our models are able to simulate a wide range of conceivable situations, the "PILACCEL" is limited in principal

Principally, such kind of direct access to the parameters and constants of models can be realized. We believe that this aspect of use of computer models will be especially interesting to provide a substitution of real nature experiments by simulation experiments. But it would require of users some special additional education in several areas (mathematics, computer technologies and programming). We think that having the required education the user could be able to change not only a limited number of parameters of loading regimes but he/she also would be able permanently to make wide the simulation situations as long as it is necessary. Only such kind of "live" model able to serve users for a long time.

RESUME

So, the basic mathematical model and appropriate software required according to the Contract has been created. The proposed model complex is able to simulate satisfactorily practically all known situations in hemodynamics of healthy men under piloting accelerations Gz. First steps have been taken towards the creation of relatively simple models for several cardiologic diseases. But to proceed in this direction we need more special experimental data for models' validation .

The proposed software has been developed mainly for experts in the area of human gravitational physiology. Besides, thanks to the intuitive user interface our software allows to use the models also as an additional illustrative material to improve the basic medico-physiological education and training of pilots. Some investigations have been done also in the direction of development of individual-oriented models that might be used in the future to find more adequate and optimal individual protection regimes for every subject. We hope that this aspect will be approved by customers for further development.

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